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Neuromechanical Analysis of Anterior Cruciate Ligament Risk Factors in Female Collegiate Soccer Athletes

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NEUROMECHANICAL ANALYSIS OF ANTERIOR CRUCIATE LIGAMENT RISK FACTORS IN FEMALE COLLEGIATE SOCCER ATHLETES

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ABSTRACT

NEUROMECHANICAL ANALYSIS OF ANTERIOR CRUCIATE LIGAMENT RISK FACTORS IN FEMALE COLLEGIATE SOCCER ATHLETES

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Old Dominion University, 2010

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The anterior cruciate ligament (ACL) acts in an essential role to prevent anterior tibial displacement when experiencing jump-landing forces that are applied to the lower extremity; more than 100,000 injuries per year in the United States in sport activities that often require landing, deceleration-acceleration, cutting and pivoting actions have been reported. The aim of this study was to examine the nature of any lower limb coupling differences between a drop-jump and a side-step cutting actions, assess kinematic and kinetic differences between three tasks, and evaluate the effects of two landing techniques in biomechanical risk factors while performing two unanticipated tasks.

Twenty female collegiate soccer athletes from a Division I institution participated in these experiments. Participants performed two unanticipated tasks; sidestep cutting and pivot, combined with two landing techniques. Three-dimensional kinematics and kinetics were recorded. The coupling relations between specific kinematic and kinetic events were assessed using principal component analysis. In addition, the degree of variability between both tasks was assessed using determination of the coefficient of variation (CV). Repeated measures analyses of variance were conducted to assess differences in the kinematic and kinetic parameters between tasks and foot landing techniques ($P<0.05$).

For experiment I, the results demonstrated that the highly loaded biomechanical variables were different between the movements, showing that the factors are inherently different depending on vertical versus horizontal oriented jump-landing tasks. Experiment II, the pivot task ($-41.2 \pm 8.8°$) had lower knee flexion and increased valgus angle (-7.6 ± 10.1) than the sidestep ($-53.9 \pm 9.4°$, and $-2.9 \pm 10.0°$, respectively) at maximum vertical ground reaction force. For experiment III, the forefoot landing technique had significantly higher knee flexion than the rearfoot ($p<0.001$), knee flexion moment ($p=0.003$), and knee adduction moment ($p<0.001$) at initial contact. This differentiation between tasks indicates that the biomechanical movement patterns are different for each movement and that the decreased variability during the drop jump may result in different adaptability of the system. During the pivot task, the athletes presented a more erect posture and adopted strategies that may place higher loads on the knee joint and increasing the strain on the ACL.

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TABLE OF CONTENTS

Page

Chapter

I. INTRODUCTION....1

Experiment I....9

Statement of Problem....9

Null Hypothesis....9

Research Hypothesis....9

Experiment II....11

Statement of Problem....11

Null Hypothesis....11

Research Hypothesis....11

Experiment III....13

Statement of Problem....13

Null Hypothesis....13

Research Hypothesis....13

Operational Definitions....15

Assumptions....17

Limitations....17

Delimitations....17

II. REVIEW OF THE LITERATURE....19

Anterior Cruciate Ligament – Anatomy & Properties....19

Epidemiology....21

Biomechanical Contributions for Anterior Cruciate Ligament Injuries....23

Hip Flexion....24

Knee Flexion....27

Knee Valgus....30

Vertical Ground Reaction Force....33

Proximal Anterior Tibial Shear Force....35

Gender....38

Landing Technique and Task....40

III. EXPERIMENT I – DIFFERENCES IN THE PATTERN OF COUPLING RELATIONS BETWEEN DROP-JUMP AND SIDESTEP CUTTING ACTIONS....46

Introduction....47

Methodology....51

Results....55

Discussion....57

Conclusions....62

IV. EXPERIMENT II – PIVOT TASK INCREASES KNEE FRONTAL PLANE LOADING WHEN COMPARED TO SIDESTEP AND DROP-JUMP....63

Introduction....64

Methodology....67

Results....70

Discussion....72

Conclusions....76

V. EXPERIMENT III – BIOMECHANICAL DIFFERENCES BETWEEN TASKS AND FOOT LANDING TECHNIQUES....78

Introduction....79

Methodology....82

Results....85

Discussion....88

Conclusions....92

VI. CONCLUSIONS....94

REFERENCES....96

APPENDICES....124

I. Figure 3.1 Typical curves of flexion angles (ankle, knee, and hip) for the drop-jump and sidestep cutting tasks....124

II. Table 3.1 Eigenvalues (EV), % of variance, and correlation coefficients between the variables for principal component 1 for the drop-jump and sidestep cutting....125

III. Table 3.2 Eigenvalues (EV), % of variance, and correlation coefficients between the variables for principal component 2 for the drop-jump and sidestep cutting 126

IV. Figure 3.2 Coefficient of variation between a drop-jump and sidestep cutting for hip, knee, and ankle flexion. Significant level at $P<0.05$ (*)....................127

V. Figure 3.3 Coefficient of variation between a drop-jump and sidestep cutting for knee valgus, hip abduction, and trunk flexion....................128

VI. Figure 4.1 Unanticipated scenario projected onto a screen to mimic a soccer game situation....................129

VII. Table 4.1 Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinematic variables at initial contact, peak stance, and peak vertical ground reaction force (PVGRF) during three athletic tasks....................130

VIII. Figure 4.2 Knee flexion angle during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks....................131

IX. Figure 4.3 Knee valgus angle during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks....................132

X. Table 4.2 Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinetic variables at initial contact and peak stance during three athletic tasks....................133

XI. Figure 4.4 Knee varus-valgus moment during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks....................134

XII. Figure 4.5 Posterior Ground Reaction Force during the three landing tasks measured in multiple of bodyweight while performing three tasks – drop-jump, sidestep, and pivot..135

XIII. Table 5.1 Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinematic variables at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PPGRF) and peak stance during two athletic tasks using two landing techniques............................136

XIV. Table 5.2 Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinetic variables at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PPGRF) and peak stance during two athletic tasks using two landing techniques............................137

XV. Figure 5.1 Hip flexion angular displacement during two tasks performed with two landing techniques throughout the stance phase. Measured in degrees...138

XVI. Figure 5.2 Knee flexion angular displacement during two tasks performed with two landing techniques throughout the stance phase. Measured in degrees...139

XVII. Figure 5.3 Knee valgus angular displacement during two tasks performed with two landing techniques throughout the stance phase. Measured in degrees...140

XVIII. Figure 5.4 Posterior Ground Reaction Force during two tasks performed with two landing techniques throughout the stance phase. Measured in multiples of bodyweight...141

XIX. Figure 5.5 Knee adduction-abduction moment during two tasks performed with two landing techniques throughout the stance phase. Measured in Nm/Kgm....142

VITA....143

CHAPTER I

INTRODUCTION

Rather than being viewed simply as noise (Harris & Wolpert, 1998; Schmidt, Zelaznik, Hawkins, Frank, & Quinn, 1979), variability of motion has been seen as an inherent characteristic of the motor system and movement performance (Newell & Corcos, 1993b). Indeed, for many voluntary actions, the presence of increased variability has been seen to be beneficial to movement performance in that it affords the individual the capacity to respond optimally to different task challenges, and subsequently reducing the likelihood of potential injuries (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Holt, Obusek, & Fonseca, 1996; Neuringer, 2002; Neuringer, 2004; Newell & Corcos, 1993a; Newell & Slifkin, 1998a; Yates, 1987). The use of variability measures to assess movement performance has been shown to be particularly useful in a variety of contexts. For example, changes in the variability of motion have been shown to discriminate between individuals on the basis of injury (Hamill, et al., 1999; Heiderscheit, 2000; Heiderscheit, Hamill, & van Emmerik, 2002b), gender (Barrett, Noordegraaf, & Morrison, 2008), neurological disorders (Dingwell & Cusumano, 2000; Dingwell, et al., 1999; Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998; Hausdorff, et al., 1997), and normal ageing (Hausdorff, et al., 1996; Hausdorff, Rios, & Edelberg, 2001).

The assessment of changes in movement variability has proven to be particularly useful for assessing adaptation to or risk of injury (Hamill, et al., 1999; Heiderscheit, 2000; Heiderscheit, et al., 2002b; Pollard, Davis, & Hamill, 2004a; Pollard, Sigward, & Powers, 2007). Hamill and colleagues (1999) reported that individuals with unilateral patella-femoral pain exhibit low variability in joint coupling during rapid, cutting maneuvers. It was subsequently argued that, because of this loss of variability, these individuals may have a reduced ability to adjust to the task demands, an outcome which potentially places them at greater risk of injury (Hamill, et al., 1999). A similar result was reported by Pollard et al. (2005) where females exhibited lower variability in lower limb joint coupling during an unanticipated cutting maneuver. This diminished variability was argued to represent a risk factor for injury because of greater localized mechanical stress on anatomical structures that may contribute in the longer term to degenerative changes from overuse (Pollard, Heiderscheit, van Emmerik, & Hamill, 2005a). Both the studies by Hamill (1999) and Pollard (2005) where designed to assess the impact of different running tasks (cutting maneuvers) on lower limb injury. One common theme of this research is to identify those factors that could contribute to damage the anterior cruciate ligament (ACL), one of the most debilitating knee ligament injuries in the collegiate athletic population (Agel, Arendt, & Bershadsky, 2005; Arendt & Dick, 1995).

The anterior cruciate ligament acts in an essential role to prevent anterior tibial displacement when experiencing jump-landing forces that are applied to the lower extremity. ACL injury has been reported to account for more than 100,000 injuries per year in the United States in sport activities that often require landing, deceleration-acceleration, cutting and pivoting actions, such as seen in soccer, volleyball, and basketball (Agel, et al., 2005; Arendt & Dick, 1995; Arendt, Agel, & Dick, 1999; Griffin, et al., 2000). ACL tears often require surgical repair, treatment and rehabilitation, which can be extremely expensive with an approximate total cost of 1.7 billion dollars per year (Griffin, et al., 2000). Knee joint osteoarthritis, increased risk of further injury, reduced sports participation, and increased laxity are examples of long-term health consequences of reconstructive ACL surgery (Gelber, et al., 2000; Lohmander, Ostenberg, Englund, & Roos, 2004; Louboutin, et al., 2008). The most common mechanism reported for ACL injury has been during a non-contact situation (Arendt, et al., 1999; Boden, Dean, Feagin, & Garrett, 2000b; Engstrom, Johansson, & Tornkvist, 1991). Strategies such as suboptimal landing (i.e., small knee flexion angle, knee valgus position), combined with high impact forces, impairments in dynamic postural control, and lower extremity strength deficits (i.e., hamstring/quadriceps ratio) have been related to an increased likelihood for injury (Ahmad, et al., 2006; Chappell, Yu, Kirkendall, & Garrett, 2002b; Ford, Myer, & Hewett, 2003; Ford, Myer, Toms, & Hewett, 2005; Hewett, Ford, Myer, Wanstrath, & Scheper, 2006; Myer, et al., 2009; Yu, Lin, & Garrett, 2006). These neuromechanical patterns have been reported during a drop-jump, sidestep cutting, or running-stop maneuver (Boden, et al., 2000b; Griffin, et al., 2006; Shimokochi & Shultz, 2008; Yu & Garrett, 2007). Each of these movements are essential to the game of soccer.

Numerous biomechanical factors across different joints and actions have been identified as potential markers for ACL injury (Blackburn & Padua, 2008; Ford, et al., 2003; Houck, 2003; McLean, Huang, Su, & Van Den Bogert, 2004; Sell, et al., 2007; Yu & Garrett, 2007). As an example, some of the theorized risk factor markers include the knee joint (decreased knee flexion, increased knee valgus angle and moments), the hip (decreased hip flexion, increased hip rotation), the tibia (increased proximal anterior tibia shear force) and the lower limb as a whole (increased peak vertical and posterior ground reaction forces) (Blackburn & Padua, 2008; Ford, et al., 2003; Houck, 2003; McLean, Huang, et al., 2004; Sell, et al., 2007; Yu & Garrett, 2007).

Compounding the problem of clearly identifying biomechanical risk factors for ACL injury, there is also a high likelihood that these risk factors change as a function of the specific task being performed and the movements that emerge (Newell & Corcos, 1993a; Newell & Slifkin, 1998a). In this regard, the movement output reflects and is affected by the specific task parameters of the action itself (task dependent factors). For ACL injury, the predictive variables that can be assessed alter as a function of the movement being performed, that is, whether the resultant action involves horizontal deceleration, vertical deceleration, and/or rotation. For many predictive injury studies, two common movements have been used to assess ACL risk factors, namely sidestep cutting (McLean, Huang, et al., 2004; McLean, Huang, & van den Bogert, 2008; Pollard, et al., 2007; Sigward & Powers, 2006a), and the drop-jump (Chappell & Limpisvasti, 2008; Cortes, et al., 2007a; Ford, et al., 2003; Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005). Arguably, the sidestepping task has been more commonly employed because of its close association with real-life athletic tasks (McLean, Huang, et al., 2004; McLean, et al., 2008; Pollard, et al., 2007; Powers, Sigward, Ota, & Pelley, 2004). The drop jump task has been utilized primarily because its landing control is comparable with other athletic tasks (e.g., sidestep cutting task), and from the experimenters' perspective this task is easier to perform under controlled laboratory settings (Noyes, Barber-Westin, Fleckenstein, Walsh, & West, 2005; Yu & Garrett, 2007). The drop jump task has also been utilized since several ACL injuries have been linked with a vertical drop landing from a jump, similar to the movement performed during a basketball rebound. Additionally, the drop-jump has been utilized as part of screening tool (Landing Error Scoring System) to evaluate individuals with potentially faulty motion patterns.

Despite the risk factors being studied under different movement tasks, there are some inherent differences in the movements that need to be considered. For example, sidestep cutting contains a significant horizontal velocity and rotational component of the segments (i.e., internal rotation of the knee) due to the change in direction (i.e., 45 degree angle); two features that are not present in the drop-jump task. Furthermore, the drop-jump entails a vertical drop from a box with minimal-to-no rotational component of the lower limb segments, whereas the sidestep cutting includes a deceleration/acceleration phase and rotational component often seen during the event of ACL tears. Given these intrinsic task differences, there is little wonder that a variety of different potential biomechanical variables across multiple lower extremity joints have been identified as risk factors for ACL injury. Nevertheless, despite the numerous factors identified, their occurrence over multiple locations within the lower limb, and the task dependent nature of the injuries, most studies have focused on reporting single risk factors as the leading cause of ACL injury (McLean, Huang, et al., 2004; Yu & Garrett, 2007). In order to gain a clearer understanding of the mechanisms of ACL injury, it is essential to identify what the risk factors are, how the different factors are actively related or coupled and whether differences in these coupling relations can be observed across different tasks.

Greig (2009) argued that the sidestep cutting does not replicate the demands of a pivot task that normally occurs during a soccer game (Greig, 2009). A pivot task, with 180 degrees of change in direction, was reported to provide a more realistic representation of a soccer task (Greig, 2009). This 180-degree maneuver commonly seen in soccer requires a complete deceleration with a change in direction followed by acceleration to maximum speed. These inherent differences suggest that the control mechanism and demands between these movements are distinctive, and the multiple biomechanical risk factors may have a different role depending on the task (Newell & Slifkin, 1998a).

Few studies have attempted to quantify and compare biomechanical parameters among tasks. The understanding of how the hypothesized risk factors act under different task constraints might provide better insight into augmented risk motions. The problematic nature of the intrinsic difference in the control mechanisms of various tasks, combined with the factor of how these tasks are conducted under laboratory experiments has been of recent concern. Biomechanical studies have focused on creating a more realistic approach through the use of light stimulus to produce the unanticipated factor (Beaulieu, Lamontagne, & Xu, 2008; Ford, et al., 2005; Pollard, Heiderscheit, Davis, & Hamill, 2004; Pollard, et al., 2005a). The light stimuli do not truly mimic a game situation, although an improvement over standard laboratory setting the environment that players normally experience is still not present under this situation. Consequently, it is essential that more realistic scenarios are developed and ultimately utilized when evaluating biomechanical parameters related to ACL risk factors. This approach to a real-life situation attempts to improve a study's ecological validity, which is often underestimated and undervalued (Robins, Hunyadi, & Schultz, 2008; Shiffman, Stone, & Hufford, 2008). The applicability and generalization of any study to real-world situations is dependent on its design (Robins, et al., 2008; Shiffman, et al., 2008). Neuropsychologists have started to focus on this factor, and have investigated the effect of conducting the studies under real-life situations (Chaytor, Schmitter-Edgecombe, & Burr, 2006; Chaytor, Temkin, Machamer, & Dikmen, 2007). Recently, Parsons and colleagues have implemented a virtual reality environment to study neurocognitive functions, which has shown to improve its reliability and (ecological) validity (Parsons, Silva, Pair, & Rizzo, 2008).

Associated with a study's ecological validity and various tasks used are the different landing techniques that an athlete can use during a landing task. Improper technique during jump-landing maneuvers may place significant force and strain on the ACL, hence causing the ligament to rupture. A few investigators have suggested that landing with a toe-to-heel (forefoot) pattern reduces landing forces and results in "safe" landing patterns (Hewett, 2000; Onate, et al., 2005; Onate, Guskiewicz, & Sullivan, 2001), however these recommendations were not fully supported by evidence. Cortes and colleagues (2007) have investigated the effects of different foot-landing techniques on knee kinematics and gender differences. The heel-to-toe (rearfoot) landing technique, oftentimes used when stopping or pivoting (e.g., basketball stop-jump, soccer pivot), was reported to present kinematic characteristics that may lead to potentially injurious landing mechanics as compared to forefoot landing strategies. In a recent 2-D video analysis of ACL injury episodes, it was suggested that at the time of injury the athletes presented a hindfoot landing (rear of the foot) (Boden, Torg, Knowles, & Hewett, 2009). The authors theorized that this landing technique is a potential mechanism to increased risk of injury, since the gastrocnemius-soleus complex cannot act and absorb the force from landing (Boden, et al., 2009). This force is directly transmitted to the knee, and most likely increases the strain at the anterior cruciate ligament. However, the study was based on 2-D video analysis without standardized distance to the field, video camera, and other essential factors for such analysis. This method to assess kinematic measures of different maneuvers has been shown to have poor reliability among experienced researchers (Krosshaug, et al., 2007a), and does not allow the assessment of ground reaction forces and consequently joint loads (kinetics). The impact forces produced from ground contact have been extensively studied as possible causes of lower extremity injury (Bisseling & Hof, 2006; Caster & Bates, 1995; Chappell, Yu, Kirkendall, & Garrett, 2002a; Coventry, O'Connor K, Hart, Earl, & Ebersole, 2006; Decker, Torry, Wyland, Sterett, & Steadman, 2003; DeVita & Skelly, 1992a; Irmischer, et al., 2004; James, Bates, & Dufek, 2003; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001b; McNitt-Gray, 1993a; Mizrahi & Susak, 1982; Zhang, Bates, & Dufek, 2000a). The injury to the lower extremity can be a consequence of numerous reasons: increased landing height, where greater kinetic energy is acquired; running speed, which creates greater displacement of the center of mass and a sudden deceleration is required to stop it; and sport task, the specificity of the task places different impact forces demands on the kinetic chain (i.e., drop jump, sidestep cutting, running stop).

Experiment I

Statement of Problem

The purpose of this study is to identify kinematic and/or kinetic variables that are highly correlated with two tasks (drop jump and sidestep cutting) using principal component analysis. We aim to identify the kinematic and kinetic variables that are highly correlated with each task.

Null Hypothesis

There will be no statistically significant differences in the coefficient of variation of kinematic variables between the drop-jump and sidestep cutting tasks.

Research Hypothesis

It is expected that the kinematic descriptors of a sidestep cutting task will be different than those of a drop-jump task. The sidestep cutting will present higher coefficient of variation in selected kinematic variables (ankle flexion, knee flexion, and hip flexion) than the drop-jump task.

The independent variables in this study will be

Movement task (drop box and side step cutting);

The dependent variables in this study will be

Kinematic variables:

Knee Flexion Angles (°)

Knee Rotation Angles (°)

Knee Valgus (°)

Ankle Flexion Angles (°)

Hip Flexion Angles (°)

Hip Rotation (°)

Kinetic variables:

Vertical ground reaction forces (Mbw)

Posterior ground reaction force (Mbw)

Proximal tibial anterior shear force (Mbw)

The kinematic and kinetic variables will be analyzed at different time instants:

Initial foot contact

Peak knee flexion

Peak vertical ground reaction force

Peak posterior ground reaction force

Peak proximal tibial anterior shear force

Coefficient of variation:

Ankle flexion

Hip flexion

Hip abduction

Knee flexion

Knee valgus

Trunk flexion

Experiment II

Statement of Problem

The purpose of this study is to analyze kinematic and kinetic strategies between tasks (drop jump, sidestep cutting and pivot). The study aims to quantify kinematic data (knee flexion, knee valgus, hip flexion, and ankle flexion) and kinetic data (vertical and posterior ground reaction forces, and knee valgus moment) between three jump-landing movement tasks.

Null Hypothesis

There will be no statistically significant difference between jump-landing movement tasks (drop jump, sidestep cutting, and pivot) in kinematic (knee flexion, knee valgus, hip flexion, and ankle flexion) and kinetic data (vertical and posterior ground reaction forces, and knee flexion and valgus moments).

Research Hypothesis

The pivot task will produce significantly lower knee and hip flexion angles, higher knee valgus angle, higher vertical and posterior ground reaction forces, higher knee flexion and valgus moments at initial foot contact, higher peak knee flexion angles, and higher peak vertical ground reaction force than the sidestep task.

The independent variables in this study will be:

Jump-landing movement tasks (drop-jump, sidestep cutting and pivot task)

The dependent variables in this study will be:

Kinematic variables:

Knee Flexion Angles (°)

Knee Valgus (°)

Hip Flexion Angles (°)

Ankle Flexion (°)

Kinetic variables:

Vertical ground reaction forces (Mbw)

Posterior ground reaction force (Mbw)

Knee valgus moment (Nm/kg)

Knee flexion moment (Nm/kg)

The kinematic and kinetic variables will be analyzed at different time instants:

Initial foot contact

Peak vertical ground reaction force

Peak posterior ground reaction force

Peak knee flexion angle

Experiment III

Statement of Problem

The purpose of this study is to analyze lower extremity motion patterns, during two-foot position aspects (forefoot, and rearfoot) while performing two tasks (sidestep cutting and pivot). The study aims to quantify the kinematic data (knee flexion, knee valgus, hip flexion, femoral rotation) and the kinetic data (vertical and ground reaction forces, and knee flexion and valgus moments) in order to evaluate the lower extremity motion pattern between the different foot-landing techniques and tasks in female collegiate soccer athletes.

Null Hypothesis

There will be no statistically significant differences between the foot landing techniques (forefoot and rearfoot) and tasks (sidestep cutting and pivot) in kinematic and kinetic variables at all time instants while performing a sidestep cutting task.

Research Hypothesis

The rearfoot landing technique will produce significantly lower knee flexion, and hip flexion, and higher knee valgus, vertical and posterior ground reaction forces, and knee flexion and valgus moments at initial foot contact, peak knee flexion, peak vertical ground reaction force than the forefoot landing technique.

The pivot task will produce significantly lower knee flexion, and hip flexion, and higher knee valgus, vertical and posterior ground reaction forces, and knee flexion and valgus moments at initial contact, peak knee flexion, peak vertical ground reaction force than the sidestep task.

The independent variables in this study will be:

Foot-landing techniques: forefoot, and rearfoot

Tasks: Sidestep cutting, and pivot

The dependent variables in this study will be:

Kinematic variables:

Knee Flexion Angles (°)

Hip Flexion Angles (°)

Knee Valgus (°)

Ankle Flexion Angles (°)

Kinetic variables:

Vertical ground reaction forces (Mbw)

Posterior ground reaction force (Mbw)

Knee flexion momen (Nm/kg.m)

Knee valgus moments (Nm/kg.m)

The kinematic and kinetic variables will be analyzed at different time instants:

Initial foot contact

Peak knee flexion

Peak vertical ground reaction force

Peak posterior ground reaction force

Operational Definitions

- Peak vertical ground reaction force is the highest data value of the vertical ground reaction forces between the initial contact and maximum knee flexion;
- Peak posterior ground reaction force is the highest data value of the posterior ground reaction forces between the initial contact and maximum knee flexion;
- Initial contact is defined as the moment where vertical ground reaction force is higher than 10 Newtons;
- Maximum knee flexion angle is the instant where the knee achieves the highest angle while the foot is in contact with the force plates. It is also the instant that defines the end of the stop-jump phase;
- Range of Motion (RoM) it is measured from the angle at initial contact until the angle at maximum flexion of a specific joint within the stop-jump phase;
- The dominant leg will be defined as the leg that the subject would use to kick a soccer ball as far as possible, and will be the tested leg during the athletic task (Ford, et al., 2003; Hewett, Myer, Ford, et al., 2005);
- Stop-jump phase is the phase that goes from initial foot contact with the force plates to the maximum knee flexion angle. It is the time where the subjects are absorbing the energy and forces from impact;

- Forefoot landing consists of initial contact with toes first on the force plates, followed by the rearfoot (FF);
- Rearfoot landing consists of initial contact with heels first on the force plates, followed by the forefoot (RF);
- Running-pivoting task is a task where the subjects will run along the platform until planting onto the force plates with the dominant foot and then pivot 180 degrees to run in the opposite direction (Greig, 2009);
- Running-stop task is a task where the subjects will run along the platform until stopping on the force plates, one foot on each plate, followed by a jump into the air as high as they can (Chappell, Creighton, Giuliani, Yu, & Garrett, 2007; Chappell & Limpisvasti, 2008; Chappell, et al., 2002b);
- Running sidestep cutting task is a change of direction to the contra-lateral side of the foot touching the force plate at an angle of approximately 45° (Colby, et al., 2000; McLean, Huang, et al., 2004; McLean, Huang, & van den Bogert, 2005; Pollard, et al., 2007).

Assumptions

- The high-speed cameras, Model MX-F40 (Vicon Motion Systems Ltd., Oxford, England) and two Bertec Force Plates, Model 4060-NC (Bertec Corporation, Columbus OH, USA), will be accurately calibrated for each subject throughout the experiments;
- The subjects will perform the protocols as asked by the researcher.

Limitations

- The subjects used for this study will be limited to female collegiate soccer athletes;
- Each subject will have varying degrees of sport experience and years of experience that may influence their jump-landing skills;
- The measurements will be done in a laboratory setting, and not in a "real life" situation (i.e., practice situation, game, etc.);
- The amount of pronation occurring at the subtalar joint will be different between subjects.

Delimitations

- This study will use collegiate female soccer athletes that must have been exercising at least 30 minutes per day, 3 times per week, for the past 6 months;
- The subjects will range in age from 18-29 years of age;

- Any subject who has had any type of surgery resulting in missing one or more days of participating, in the lower extremity, ankle and knees, within the previous twenty-four months will be excluded from this study (Onate, et al., 2005);
- Any subject that reports any physical impairment that will limit them in performing a jump-landing task will be excluded from this study;
- Any subject who is presently pregnant will be excluded from this study;
- All the subjects will be using the same model of sneakers to control for possible shock-absorption differences between shoes.

CHAPTER II

REVIEW OF THE LITERATURE

The following review of literature will focus on the anterior cruciate ligament and the associated concerns: epidemiology, risk factors for injury, jump-landing tasks used, and foot-landing techniques, among others. These aspects have been theorized as important factors in trying to understand the underlying mechanism of injury. While many studies have focused on analyzing single risk factors of anterior cruciate ligament injury, there is a need to better understand how these risk factors correlate among different tasks, and within various foot-landing techniques.

Anterior Cruciate Ligament – Anatomy & Properties

The anteromedial and posterolateral bundles are two functional bundles of the ACL (Petersen & Zantop, 2007). The cruciate ligament has a length of 38 mm, ranging from 25 to 41 mm, a width of 10 mm, ranging from 7 mm to 12 mm, and a volume of 2.3 ± 0.4 mm (Kennedy, Weinberg, & Wilson, 1974; Odensten & Gillquist, 1985). This ligament is made up of multiple collagen fascicles, which are surrounded by an endotendinuem (Kennedy, et al., 1974; Odensten & Gillquist, 1985). These collagen fascicular units measure 0.25 to 3 mm in diameter (Amis & Dawkins, 1991). The major blood supply is provided by the middle genicular artery, as the bony attachments of the anterior cruciate ligament do not provide a significant source of blood (Arnoczky, Rubin, & Marshall, 1979). The ACL attaches to the femur and the tibia. The femoral attachment is oriented in the longitudinal axis, starting at the posteriormedial corner of the medial aspect of the lateral femoral condyle in the intercondylar notch (Amis & Dawkins, 1991; Petersen & Zantop, 2007). The tibial attachment is located in the anteroposterior axis of the tibia, and attaches in front and lateral of anterior spine inserting into the interspinous area of the tibia (Amis & Dawkins, 1991; Petersen & Zantop, 2007).

The ACL has an ultimate tensile load of 2160 ± 157 N and a stiffness of 242 ± 28 N/mm (Woo, Hollis, Adams, Lyon, & Takai, 1991). At knee extension, the posterolateral bundle of the ACL is tight while the anteromedial bundle is somewhat relaxed (Amis & Dawkins, 1991; Petersen & Zantop, 2007; Woo, et al., 1991). This extension and hyper-extension causes greater stress in the ACL than in the posterior cruciate ligament (PCL) (Amis & Dawkins, 1991; Petersen & Zantop, 2007). In knee flexion, since the femoral attachment of the ACL moves into a more horizontal position, the anteriormedial bundle tightens, while the posteriolateral bundle relaxes (Amis & Dawkins, 1991; Kennedy, et al., 1974; Odensten & Gillquist, 1985; Petersen & Zantop, 2007; Woo, et al., 1991). At 40 to 50 degrees of knee flexion the tension is minimal in the ACL (Kennedy, et al., 1974). There is a known relationship between quadriceps muscle activity and anterior cruciate ligament strain at different knee flexion angles. At 30 degrees of flexion, the ACL is significantly higher strained than at 90 degrees where the ligament does not experience strain (Amis & Dawkins, 1991; Markolf, Wascher, & Finerman, 1993; Wascher, Markolf, Shapiro, & Finerman, 1993; Woo, et al., 1991).

EPIDEMIOLOGY

Anterior cruciate ligament (ACL) failure accounts for approximately 100,000 injuries in the United States (Arendt & Dick, 1995; Griffin, et al., 2000; Griffin, et al., 2006; McLean, Huang, et al., 2004; McLean, Su, & van den Bogert, 2003; McLean, Walker, & van den Bogert, 2005). The rehabilitation process from such an injury can be long and agonizing; oftentimes requiring surgical reconstruction (McLean, Walker, et al., 2005). Roughly 50,000 injuries end up requiring surgical reconstruction (Griffin, et al., 2000; McLean, Huang, et al., 2004; McLean, Walker, et al., 2005). The surgical procedure and associated rehabilitation has an estimated cost of almost two billion dollars per year (Griffin, et al., 2000). This cost does not account for the long-term consequences, such as knee osteoarthritis (OA) and all problems associated with that. It is well reported that knee OA tends to develop in subjects that experienced ACL injury. The suggested rates for knee OA development range from 16-90% of subjects after 5 to 15 years of experiencing the traumatic injury (Neuman, et al., 2008; Neuman, et al., 2009).

Within this tremendous amount of injuries, seventy percent are a result of non-contact situation (Arendt & Dick, 1995; Boden, Dean, Feagin, & Garrett, 2000a; Griffin, et al., 2000). This mechanism is of special interest since it is solely based on the individual characteristics (i.e., anatomatical, hormonal, neuromechanical). The non-contact ACL injury is commonly observed in female athletes, with a tremendous discrepancy in gender incidence; women are two to eight times more likely to sustain a non-contact ACL injury than men (Arendt & Dick, 1995; Griffin, et al., 2000; Griffin, et al., 2006; McLean, Huang, et al., 2004; McLean, Walker, et al., 2005; Myer, Ford, McLean, & Hewett, 2006; Russel, Palmieri, Zinder, & Ingersoll, 2006; Sigward & Powers, 2006a). A recent longitudinal study found that the number reported by Arendt and Dick in 1995 still remains valid today, with female collegiate athletes presenting a higher rate of ACL injury than males (Agel, et al., 2005).

A possible reason accounting for this gender disparity is that female athletes have been reported to utilize a neuromechanical strategy that might place them at higher risk for non-contact ACL injury than male athletes (Decker et al., 2003; Zhang et al., 2000; DeVita and Skelly, 1992; Schot et al., 1991). This strategy is directly associated with low knee flexion and lower range of motion and higher vertical ground reaction forces (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Hewett, Myer, & Ford, 2005; Hewett, Myer, Ford, et al., 2005; Lephart, Ferris, Riemann, Myers, & Fu, 2002b; Malinzak, et al., 2001b; Salci, Kentel, Heycan, Akin, & Korkusuz, 2004a). Nevertheless, some authors question the real potential of sagittal plane knee motion in ACL injuries (Hewett, et al., 2006; Hewett, Torg, & Boden, 2009; McLean, Huang, et al., 2004). It has been suggested that the sagittal plane difference between genders is directly related to the task being performed and not a risk factor *per se*. The observed differences in sagittal plane knee motion have been suggested to not be directly associated to non-contact ACL injury (McLean, Walker, et al., 2005; Myer, et al., 2006). Another reason that seems to contribute to the gender difference is the greater valgus motion that woman present. Male individuals tend to be in a varus knee position during landing from a jump (Ford, et al., 2006; Hewett, et al., 2004; McLean, Walker, et al., 2005; Myer, et al., 2006; Russel, et al., 2006). Combined with an increased knee valgus angle, is an augmented valgus force. This added loading has been related to an increase in ACL strains (Ford, et al., 2006; Hewett, et al., 2004; McLean, Walker, et al., 2005; Russel, et al., 2006). The ACL strain associated with valgus motion and loading is in concurrence with other factors, such as lack of knee flexion and weak hamstring muscles that might place women at greater risk of non-contact ACL injury. Among all the reported risk factors (i.e., hormonal, anatomical), the neuromechanical are potentially modifiable, thus focus on neuromechanical factors deserves strong attention from researchers.

Biomechanical Contributions For Anterior Cruciate Ligament Injuries

Landing and cutting maneuvers are considered high-risk motions linked to ACL injury (McLean, Walker, et al., 2005; Russel, et al., 2006; Sigward & Powers, 2006b). These maneuvers are characterized by sudden deceleration, landing and pivoting, have been frequently observed in non-contact ACL injuries (Boden, et al., 2000b). Mechanisms such as suboptimal landing strategies (small knee flexion angle, higher knee valgus angle) combined with high impact forces and sudden decelerations appear to increase the probability for injury. These mechanisms have been identified as the leading causes of ACL injuries, particularly in female athletes (Arendt & Dick, 1995; Decker, Torry, Wyland, Sterett, & Steadman, 2003; Griffin, et al., 2000; Griffin, et al., 2006; Irmischer, et al., 2004).

Previous research has further identified and classified possible risk factors related to ACL injuries (Agel, et al., 2005; Anderson, Dome, Gautam, Awh, & Rennirt, 2001; Arendt & Dick, 1995; Bobbert & Van Zandwijk, 1999; Ford, et al., 2003; Griffin, et al., 2000; Griffin, et al., 2006; Hewett, et al., 2004; Huston, Vibert, Ashton-Miller, & Wojtys, 2001; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001a; McLean, Huang, et al., 2004;

Mesfar & Shirazi-Adl, 2006; Wojtys, Huston, Lindenfeld, Hewett, & Greenfield, 1998). These risk factors have been classified as intrinsic (anatomical, hormonal, biomechanical, and neuromuscular factors) and extrinsic (playing surfaces, and equipment) (Murphy, Connolly, & Beynnon, 2003). However, the most commonly used classifications were proposed as: environmental, anatomical, hormonal, and neuromuscular (Griffin, et al., 2000; Griffin, et al., 2006).

Environmental factors are related to meteorological conditions, the type of surface, footwear and the interaction with the surface; and protective mechanisms (Griffin, et al., 2006). The anatomical risk factors have been sub-divided into *Q* angle, knee valgus, foot pronation, body mass index, notch size, ACL geometry, and ACL material properties (Griffin, et al., 2006). Hormonal risk factors were classified as anterior knee laxity and menstrual cycle phase. The neuromuscular risk factors were classified as altered movement patterns, altered muscle activation patterns, and inadequate muscle stiffness. Major contributions have been made to congregate a deeper understanding of the factors that can lead to non-contact ACL injuries, yet little progress as been achieved in explaining this multi-factorial injury. The mechanism of injury is rather puzzling, most likely due to the multitude of factors involved, which include tibial slope, knee laxity, lower extremity strength, and multiple planes of motion, just to name a few.

Hip Flexion. Analyses covering hip flexion motion have not been extensively documented in the landing literature. Only recent studies focusing on hip motion have come to the forefront of research, and it has been suggested that hip kinematics and kinetics might be of crucial contribution to non-contact ACL tear (Hewett, et al., 2006; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Pollard, Davis, & Hamill, 2004b; Pollard, Heiderscheit, van Emmerik, & Hamill, 2005b; Yu, et al., 2006; Yu, McClure, Onate, Guskiewicz, & Kirdendall, 2004; Zazulak, et al., 2005). An increase in hip flexion angles, during an event such as landing, can be useful to assist the lower extremity to absorb landing forces. Also, if there is a lack of hip flexion, the subject will land in a more erect position, possibly inducing a stiffer landing. The combination of hip kinematic and kinetic variables might place the knee in a more valgus position, which can increase the chance of the ligament collapsing (Hewett, et al., 2006). The authors have suggested that this was due to the fact that the female athletes had increased hip adduction. Nonetheless, it was suggested that an increase in hip musculature strength might avoid the excessive hip adduction and internal rotation and protect the knee from the valgus position/loading. Pollard and colleagues (2005), however, did not find any gender differences in soccer players while performing a cutting maneuver when evaluating knee and hip flexion. They suggested that sports background and years of experience are possible vital aspects to take into consideration when trying to determine risk factors for lower extremity injuries. In contrast, Yu et al. (2006) found significant differences between genders in hip kinematics and kinetics. The authors also reported that hip angular velocity has a greater influence in proximal anterior tibial shear force than knee angular velocity (Yu, et al., 2006).

It surfaces that it is necessary to analyze the lower extremity joints in an interactive way, rather than separately, since hip angular velocity appears directly related to posterior ground reaction force and proximal tibial shear force, whereas knee angular velocity has a relationship with vertical ground reaction force (Tillman, Hass, Chow, & Brunt, 2005). However, it should be noted that Yu et al. (2006) found these relationships at initial contact, when it has been stated that ACL tear is more prone to occur shortly after initial contact. Yet, the relation with knee injuries is not clearly understood and further research has been suggested (Hewett, et al., 2006; Hewett, et al., 1999; Pollard, Davis, et al., 2004b; Pollard, et al., 2005b; Yu, et al., 2006).

Many research studies that have performed lower extremity analysis have focused more on the knee and ankle motion than on the hip. In order to fully understand how the lower extremity kinetic chain absorbs the impact of landing, it is essential to include the hip in the analysis, as it is the link between the lower extremity and the upper extremity. Previous research has found that females presented less hip flexion angles, as well as lack of knee flexion (Kovacs, et al., 1999; Salci, Kentel, Heycan, Akin, & Korkusuz, 2004b; Swanik, et al., 2004). Kovacs and associates (1999) reported that using a forefoot landing strategy the subjects were able to flex the hip twofold greater than when using a heel-toe strategy. Furthermore, the hip and pelvis neuromechanical characteristics may play a vital part for the knee function as reported by Ferries et al. (2004).

A suggestion made by some researchers is that greater hip flexion angles might assist in the prevention of ACL injuries, although excessive trunk flexion, normally associated with higher hip flexion angles, might deter performance (Swanik, et al., 2004). However, mixed reviews exist, as some studies reported that females had higher hip flexion angles than their male counterparts while others found no significant difference between genders (Habu, Sell, Myers, Abt, & Lephart, 2004; Kernozek, Van Hoof, Torry, Cowley, & Tanner, 2004). When landing from any given task, subjects adopt different strategies in order to facilitate energy absorption. Such strategy is related to the use of the hip flexors, which will create hip moments to increase hip flexion. This will eventually make a softer landing by increasing the hip range of motion. If the hip moment strategy does not occur, the landing will be much stiffer, increasing the forces to be absorbed by the spine (Devita & Skelly, 1992b). Furthermore, it was reported that females use more hip strategy than males to absorb the acquired energy when landing from a forward jump, implying that female athletes use more hip extension moments whereas males use more ankle plantar flexion moments (Ford, Myer, Divine, & Hewett, 2004). Hart et al. (2004) suggested that females tend to dissipate the forces by utilizing a hip joint strategy, whereas males tend to use an ankle joint strategy (Hart, Garrison, Kerrigan, Boxer, & Ingersoll, 2004). This suggests that profound biomechanical analysis and understanding of the hip motion as part of the lower extremity kinetic chain is important, especially in female subjects. Nonetheless, some studies did not find any gender differences in hip flexion angles at its maximum value during a hopping task (Jacobs & Mattacola, 2005). Further clarification is necessary to understand the role that the hip plays in landing mechanics. A novel way to analyze it is through the use of joint coupling, based on dynamic systems theory, and comprehending the interaction between the lower extremity joints.

Knee Flexion. A risk factor frequently stated in the literature for non-contact ACL injuries is the knee flexion angle at time of injury. It has been suggested that small knee flexion angle might be associated with ACL injury (Agel, Bershadsky, & Arendt, 2006; Arendt & Dick, 1995; Boden, et al., 2000b; Griffin, et al., 2000; Griffin, et al., 2006;

Huston, Greenfield, & Wojtys, 2000; Huston, et al., 2001). The lack of knee flexion, meaning a more erect posture at initial contact, has been found to be more prevalent in women than in men (Chappell, et al., 2002b; Cowling & Steele, 2001; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Lephart, Ferris, Riemann, Myers, & Fu, 2002a; Malinzak, et al., 2001a; Salci, et al., 2004b; Trowbridge, Winder, Hunter, & Ricard, 2004). The more erect posture of females has been hypothesized as one of the primary risk factors for ACL injuries, especially when performing athletic tasks with fatigue and/or losing balance (Decker, Torry, Noonan, Riviere, & Sterett, 2002). McLean and associates (2007) reported that females had less knee flexion at initial contact than their male counterparts while under fatigue and performing a drop jump task (McLean, et al., 2007). Nyland and colleagues (1999) studied the effects of fatigued hamstrings on a crossover-cutting maneuver. At initial contact, under the non fatigue condition, the authors reported that the knee flexion value was 19 degrees (Nyland, Caborn, Shapiro, & Johnson, 1999). In comparison with the recommended angles from some authors (30 degrees of knee flexion) it can be perceived that the subjects studied by Nyland et al. (1999) might be at higher risk of injury prior to fatigue. It is noteworthy to analyze the maximum knee flexion obtained by the same subjects. With a value of 57 degrees, it implies that they had approximately 40 degrees of range of motion. This seems in accordance with recommendations that the range of motion should be higher than 30 degrees to properly absorb the impact forces. Even though the subjects were at "higher risk" at initial contact they had an optimal range of motion to decrease the potentially dangerous position.

Interestingly, some research has reported that women land with greater knee flexion than men do (Fagenbaum & Darling, 2003). Decker and associates (2003) mentioned that even though females had less knee flexion at initial contact, they presented higher range of motion at the knee, which may be a protective mechanism to dissipate the high impact forces. Arendt and Dick (1995) indicated that a flexed knee allows the knee joint to be in a more favorable position for the hamstring muscles to stabilize the joint by controlling rotation and anterior displacement. Thus, it seems that because female athletes land in a more erect position, more anterior tibial displacement may occur, placing them at a higher risk of ACL tears (Arendt & Dick, 1995; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003). Additionally, it has been shown that male athletes take more time to achieve maximum knee flexion after contact. Therefore, if females have less knee flexion and less time to reach maximum knee flexion, the lower extremity will absorb the energy more quickly which may be a factor for increased risk injury (Lephart, et al., 2002a). Jacobs and Mattacola (2005) did not find any significant difference in knee flexion angles at its peak between genders while performing a hopping task (Jacobs & Mattacola, 2005). In a study that compared lower extremity kinematics between gender while performing two tasks (running and sidestep cutting), the authors did not find any differences in knee flexion between genders in either task (McLean, Neal, Myers, & Walters, 1999). One explanation provided for the lack of difference, was that males and females at the competitive level tend to have similar running patterns. However, Sigward and Powers (2006) compared experienced and novice female subjects, while performing a sidestep cutting task and did not find any significant difference in knee kinematics (Sigward & Powers, 2006a).

In a comparison study between subjects with ACL reconstruction and non-injured, it was found that ACL injured subjects had less knee flexion at initial contact than the non-injured group. Nonetheless, the knee flexion range of motion for both groups was identical, meaning that the subjects used other strategies to compensate, such as using the hip extensors less and the ankle plantarflexors more (Decker, et al., 2002). Also, Hewett et al. (2005) when comparing female athletes with and without previous ACL injuries, found that there was no difference in knee flexion at initial contact, although there was a significant difference in maximum knee flexion, where the non-injured group had lower knee flexion angles than the injured group. This possibly means that ACL injured subjects regulate their landing strategy as a protective mechanism to the injured knee.

Knee Valgus. Higher knee valgus angle are normally associated with less knee flexion angles, especially in female subjects. Studies have investigated a variety of athletic tasks and mentioned that not only do females present less knee flexion, but they had increased knee valgus angles (Buchanan & Vardaxis, 2004; Ford, et al., 2003; Jacobs & Mattacola, 2004; Malinzak, et al., 2001a; Russell, Palmieri, Zinder, & Ingersoll, 2006; Trowbridge, et al., 2004). This is important in view of the fact that knee valgus has been cited as one biomechanical risk factor at the instant of non-contact ACL injury event (Agel, et al., 2005; Arendt & Dick, 1995; Boden, et al., 2000b; Griffin, et al., 2000; Griffin, et al., 2006; Huston, et al., 2001). However, some studies have reported that there is no gender differences in knee valgus angles (Claiborne, Armstrong, Gandhi, & Pincivero, 2006). One suggestion made by the authors was that females started already in a knee valgus position and remained in that position throughout the experiment. In contrast with the previous authors, Russel et al. (2006) found that female subjects also landed in a knee valgus position at initial contact while males landed in a varus position, although the female subjects tended to displace the knee into varus angle throughout the motion, and achieving that position at maximum knee flexion. This is relevant since the range of motion for each rotation might be an important factor to attenuate the forces as well as to reduce the stress placed on the ligaments, specifically on the ACL. Some authors have suggested that valgus angles associated with greater valgus moments can be an essential factor for the gender difference rate of ACL tears (Bendjaballah, Shirazi-Adl, & Zukor, 1997).

Ford and colleagues found that females had a significant difference in knee valgus angle and maximum knee valgus motion, but they found no difference at initial contact when compared to male subjects. They mentioned that the female subjects in this study presented ligament dominance, which is related to the inability to control the lower extremity joints by purely using the muscles (Ford, et al., 2003). Such lack of muscle control on the lower extremity might place the athletes at higher risk of ACL tear, as it has been theorized as a risk factor for this injury (Ford, et al., 2003). In a later study, Ford et al. found that youth female athletes presented higher knee valgus angles at initial contact than youth male athletes while performing an unanticipated cutting task, although, surprisingly, there was no difference at maximum knee valgus (Ford, et al., 2005). Consequently, males needed to have higher range of motion on knee valgus than females to achieve similar maximum results. This can be also a dangerous factor for ACL tears. Not only can the angle *per se* be a risk factor, but the range of motion in that biomechanical variable, can similarly contribute to higher risk of injury where a lack of range of motion may represent a lack of ability to dissipate the impact forces. In a prospective study, Hewett and colleagues have found that females with previous ACL injuries had higher knee valgus than those non-injured. The authors suggested that excessive valgus motion is linked to ACL injuries and noted that this factor should be carefully controlled in female subjects with a previous history of ACL injury (Hewett, Myer, Ford, et al., 2005). Under a hopping study, the authors did not find a statistically significant difference between genders in knee valgus, although they suggested that the difference presented might be meaningful for clinicians (females 14.30°, males 9.87°) in implementing strategies to minimize this knee valgus angle (Jacobs & Mattacola, 2005).

McLean and colleagues stated there was no gender difference in knee valgus angles in a running task in their 2007 study. However, they previously had found a significant difference between males and females when performing a sidestep cutting task, where females had higher knee valgus than males did (McLean, et al., 1999). It is noteworthy that both genders were in a knee valgus position, with males not being in varus position as expected, which lead the authors to comment that the results were not clinically significant even in the presence of a statistical difference (McLean, et al., 1999). The authors advocated that since females tend to possess higher *Q* angles, this may induce a higher moment arm promoting higher knee valgus position, which has been suggested as a risk factor for ACL tears (McLean, et al., 1999). In another recent study, the same authors reported, once again, gender differences in knee valgus across three tasks that have been associated with ACL injury (jump landing, sidestep cutting, and shuttle run) (McLean, Walker, et al., 2005). They not only found differences at initial contact, but also higher peak knee valgus, suggesting that this common factor across females can induce higher stress on the ligament causing it to collapse (McLean, Walker, et al., 2005).

Other authors have argued that knee valgus angles and moments do not appear to be a contributing factor for injury (Chappell, et al., 2002b). In a comparison between tasks and athletic population (basketball and soccer) in female athletes, the authors did not find any significant difference between sports, though there were greater knee valgus angles when the athletes performed the cutting task than when they did the landing task (Cowley, Ford, Myer, Kernozek, & Hewett, 2006). This might be due to the foot placement and task demands: athletes have to place the weight on one foot and rotate over it, which automatically places the knee in a more valgus position.

Vertical Ground Reaction Force. High vertical ground reaction forces have been associated with lack of knee flexion angles (Bobbert, Huijing, & van Ingen Schenau, 1987a, 1987b; Dufek & Bates, 1990). Especially the combination of high vertical ground reaction forces and an erected position might pose a threat for the lower extremity to collapse (Dufek & Bates, 1990). These authors have suggested that it might be possible to change the vertical ground reaction force using the appropriate landing strategy. Higher values for vertical ground reaction force have also been related to different landing techniques (Kovacs, et al., 1999; Self & Paine, 2001). Kovacs et al. (1999) found significant differences in vertical forces between heel-toe landing and forefoot landing; the vertical forces were 3.8 times higher in heel-toe landing. In the heel-toe landing the center of pressure is positioned at the calcaneus, which makes it almost impossible for the ankle plantar flexors to work properly and aid in the energy absorption at impact. Furthermore, Self and Paine (2001) found that landing on the heels presented higher vertical ground reaction forces. It has been suggested that as height increases, the vertical ground reaction force also increases (McNitt-Gray, 1993b; McNitt-Gray, 1993c; Seegmiller & McCaw, 2003; Zhang, Bates, & Dufek, 2000b), although there were some methodological differences between the studies. McNitt-Gray (1993) only studied gymnasts while landing from three different heights, whereas Seegmiller and McCaw (2003) evaluated the effect of three different heights in vertical ground reaction force between gymnasts and recreational athletes. In this later study, it was found that gymnasts had higher vertical ground reaction force when compared to the recreational group when landing from 60 cm and 90 cm. Their results might be due to the sports demands. Gymnasts are required to land as erect as possible and keep their balance in that position in a matter of milliseconds to achieve maximum score from the judges. This innate response is likely to play an important role in their landing position, more than the landing height. Nonetheless, Zhang and associates (2000) have suggested that an understanding of the relationships between landing technique and drop height is necessary, because different landing techniques might assist in reducing the vertical forces from higher landings, which ultimately might be beneficial for gymnasts and others alike.

Interestingly, some authors did not find any significant difference in vertical ground reaction force between genders (Lephart, et al., 2002a; Onate, et al., 2004), even though Lephart and colleagues (2002) found that females landed with significantly less knee flexion than males did. Thus, even though males and females presented differences in certain kinematic variables (i.e., knee flexion), they may have used other strategies to compensate and attenuate the forces that were applied on the body. In contrast, Salci et al. (2004) reported that female volleyball athletes had higher vertical ground reaction forces than their male counterparts. Mizrahi and Susak (1982) have declared that a strategy to reduce the vertical ground reaction forces and its consequences is to have a greater range of motion in the most important joints in the lower extremity (ankle, knee, and hip).

ACL injuries have been reported to have a higher incidence in soccer players than basketball (Agel, et al., 2005). Interestingly, one study stated that soccer players had less vertical ground reaction force while landing from a drop jump than basketball players (Cowley, et al., 2006). However, when these athletes had to perform a cutting task, basketball players had lower vertical ground reaction force than soccer players. It is noteworthy that in a simple task such as a drop jump, the soccer players had lower vertical ground reaction forces, but in a more demanding task (cutting) and commonly performed by those athletes, their values saw a drastic increase. Ford et al. (2003) did not find any significant difference in normalized vertical ground reaction force between genders. It was suggested that the attenuation of vertical ground reaction forces in females were caused by the intervention training, which may have altered their landing patterns and assisted in reducing the high vertical ground reaction forces to similar values as their male counterparts (Ford, et al., 2003).

Proximal anterior tibia shear force. A shear force is a force applied parallel to the surface of an object acting along the surface, creating deformation internally in an angular direction (Hamill & Knutzen, 2003; Zatsiorsky, 2002). A shear force is a result of a compression load. The knee suffers a compression load when landing from a variety of athletic tasks, resulting in an anterior-posterior shear force. The knee acts as a lever between two of the major bones in the human body - the femur and tibia, and thus significantly greater loadings at the knee joints are expected (Zhang, et al., 2000b). Those increased loads at the knee joint may place the knee at higher risk of injury (Zhang, et al., 2000b). Thus, it seems pertinent to quantify the forces and moments that are applied on this structure, since it is an essential and sensitive lever used for human locomotion and oftentimes with ligamentous injuries. It is important to understand the amount of load that arises in the knee joint, especially the proximal tibia anterior shear force, after different landing techniques and athletic tasks. It has been suggested by Chappell et al. (2002) that the proximal tibia anterior shear force can cause ACL injuries, as the proximal tibia anterior shear force can bring the tibia forward and away from the femur, thus rupturing the lever between these two. The authors assert that such tension is directly related to tearing of the ACL. Due to the horizontal displacement, different tasks and landing techniques will induce different proximal tibia anterior shear forces during landing and is therefore an important are for further study.

Gender comparison studies have reported that female athletes, when compared to male athletes, present significantly higher proximal tibia anterior shear forces (Sander, et al., 2004; Sell, et al., 2004). This can be related to greater use of quadriceps instead of hamstring muscles to prevent anterior displacement of the tibia. Knee laxity seems to play an important role on ACL injuries, with less knee laxity being referenced as a possible cause to help prevent the tibial anterior shear forces and displacement (Rozzi, Lephart, Gear, & Fu, 1999; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001). It has been found that non-athletes have significantly higher knee laxity than athletes do, thus potentially placing them at higher risk due to the possible higher strains that is experienced by the ACL (Bowerman, Smith, Carlson, & King, 2006). Surprisingly, the authors did not find any difference between genders (Bowerman, et al., 2006). In contrast, Rozzi et al. (1999) revealed that female athletes had higher knee laxity than male athletes do, and this can be one reason for higher rates of non-contact ACL injuries in females when experiencing tibial anterior shear forces. Sell and colleagues (2006) studied a stop-jump task under two conditions – anticipated and unanticipated, and tried to understand its effect on joint kinematics and kinetics between gender (Sell, et al., 2006). The authors found that female subjects had significantly higher proximal tibia anterior shear force than males did. It was argued that an association between decreased knee flexion angle and increased shear force, increasing the anterior tibia translation on females might have placed more strain on the ACL (Sell, et al., 2006). More recently, Sell and his associates (2007) have reported predictor variables of proximal tibia anterior shear force during a vertical stop-jump. The authors reported that the strongest and most significant predictors of proximal tibia anterior shear force were those occurring at peak posterior ground reaction force (Sell, et al., 2007). A negative correlation between knee flexion moment and proximal tibia anterior shear force seemed to be the strongest one, followed by a positive correlation between knee flexion angle and proximal tibia anterior shear force, and between shear force and posterior ground reaction force.

Gender. Gender has been suggested as a major risk factor for non-contact ACL injury, where women are two to eight times more likely to sustain an ACL injury than men (Arendt & Dick, 1995; Griffin, et al., 2000; Griffin, et al., 2006; McLean, Huang, et al., 2004; McLean, Walker, et al., 2005; Myer, et al., 2006; Russel, et al., 2006; Sigward & Powers, 2006a). A longitudinal study performed between 1990 and 2002 reported that female collegiate athletes (basketball and soccer) have a higher rate of ACL injury than males do (Agel, et al., 2005). There was a noticeable decrease of ACL injuries in male athletes from 1990-2002. Despite the decrease, there was a significant difference between soccer and basketball sports, with a higher rate of injuries in soccer players (Agel, et al., 2005). The disparity between genders in non-contact ACL injury obliges us to further understand the biomechanical factors behind it. It seems pertinent to evaluate soccer female characteristics, as they are one of the most commonly impaired athletic populations as a result of non-contact ACL injuries.

Previous research has presented evidence that females exhibit different landing patterns than males do (Chappell, et al., 2002b; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Fagenbaum & Darling, 2003; Huston, et al., 2001; Jacobs & Mattacola, 2004; Lephart, et al., 2002a; Malinzak, et al., 2001a; Salci, et al., 2004b). The previously found differences mainly occur in the knee and hip joint, where females tend to be in a more erect position, with less knee flexion and hip flexion associated with excessive knee valgus angles (Chappell, et al., 2002b; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Fagenbaum & Darling, 2003; Huston, et al., 2001; Jacobs & Mattacola, 2004; Lephart, et al., 2002a; Malinzak, et al., 2001a; Salci, et al., 2004b). Ford and associates found that females present a higher excursion in hip and knee motion, and suggested that this factor can be associated with higher risk for ACL injuries (Ford, et al., 2006). In a recently published study, Weinhold et al. (2007) found that when applying experimental loading patterns to knee cadavers, females placed a higher load on the ACL strain than did males (Weinhold, et al., 2007). Pollard et al. (2004) did not find any significant differences between males and females in a gender comparison during a randomly cued maneuver. The authors investigated the effect of three randomly cued maneuvers in the lower extremity, hip and knee joint. One of the reasons to account for the lack of gender differences is the originality of the task, which may be responsible for the difference in knee abduction angles when compared with the ones reported by Malinzak et al. (2001) and McLean et al. (1999). Nonetheless, both genders presented kinematic and kinetic data theorized to place them at higher risk for non-contact ACL tears. Sigward and Powers (2006) did not find any significant difference between males and females in knee kinematics (flexion, abduction, and rotation). One reason provided by the authors for the lack of gender differences was the small sample size causing high variability in the knee kinematic variables analyzed (Sigward & Powers, 2006b). More recently, the same authors (Pollard, et al., 2007), in a gender difference study during sidestep cutting maneuvers, found that female athletes had different hip characteristics than their male counterparts, with the female athletes exhibiting greater hip internal rotation. The authors recommended that it might be necessary to evaluate strength differences between males and females in future studies to determine if the obtained difference could be due to a strength deficit in the hip musculature.

Pollard and colleagues (2005) when evaluating lower extremity joint coupling study that females had less coupling variability than males (Pollard, et al., 2005a). The authors studied the coupling of lower extremity joints in several of rotations (flexion, rotation and abduction), using the same rotation between joints as well as using different rotations between and within joints. The lack of coupling variability, as suggested by the authors, may mean that females tend to lack adaptation when facing perturbations in the neuromuscular system. As a result, this lack of adaptation was suggested as one reason for the higher incidence of non-contact ACL injuries in females as compared to males. Other studies similarly have proposed that low coupling variability may be associated with lower extremity injuries (Hamill, et al., 1999; Heiderscheit, Hamill, & van Emmerik, 2002a). Although it can be argued that high coupling variability might also place the subjects at high risk due to the lack of consistency within the neuromuscular system in a cutting maneuver for experienced soccer athletes.

A few authors found that females landed with greater knee flexion than males, and suggested that the difference in the injuries that occurred during landing activities may not be related to gender, but rather to landing patterns (Fagenbaum & Darling, 2003; Pollard, et al., 2005a; Swartz, Decoster, Russel, & Croce, 2005). The contradictions in the literature require further investigation of the biomechanical factors in female athletes while performing specific athletic tasks that are associated with high incidence of non-contact ACL injury. Lastly, the line of research for joint coupling might be beneficial for females and should be studied in isolation (Tillman, et al., 2005).

Landing Technique and Task. Different landing techniques can induce different amounts of loads on the lower extremity, which may in turn lead to injuries. According to Butler et al. (2003) many studies showed that knee stiffness is related to landing on the toes, while ankle stiffness is related to heel landings. These findings suggest that different landing patterns will affect how the body absorbs the energy and its forces. It is necessary to obtain a better understanding of the mechanisms underlying the different landing techniques and how the lower extremity joints act during various landing types (Tillman, et al., 2005). It has been suggested that various landing techniques produce different absorption force and energy patterns that can be dangerous for the lower extremity (Butler, Crowell III, & Davis, 2003). Surprisingly, in a comparison study between landing techniques, no significant difference was found in knee flexion between heel-toe and toe-heel landing (Kovacs, et al., 1999).

Different landing techniques change the motor organization of the subjects, with a special emphasis on the neuro-musculoskeletal requirements (Schot & Dufek, 1993). The neuro-musculoskeletal organization may place the individuals at higher risk in situations where they have to use a technique that they are unfamiliar with. The factor that has been mentioned has poor landing technique, which can be linked to ACL injuries during dynamic activities (Cowling & Steele, 2001). It has been suggested that vertical ground reaction forces can be reduced using a toe-heel landing strategy when compared to other techniques (i.e., flatfoot, heel-toe, etc.), as well as that different motion patterns create less variability in the subjects' kinematic and kinetic variables (Dufek & Bates, 1990; Self & Paine, 2001). Hence, it is pertinent to understand how different landing techniques influence the subjects' motion during various athletic tasks.

In a recently published study, Cortes and associates (2007) did not find a significant difference between genders, however they did report a significant difference between landing techniques, with a rearfoot technique presenting values that might place the athletes at higher risk of injury. Nonetheless, the task used in the study was a simple non-athletic drop jump, and it is necessary to employ the same approach with more realistic athletic tasks, such as running stop, sidestep cutting, pivoting, etc. The approach taken in the study by Cortes et al. (2007) is that the lower extremity is analyzed from a distal to proximal approach (ankle – knee – hip), instead of the traditional approach of proximal to distal joints (hip – knee – ankle). The heel-to-toe landing strategy may place greater stress at the knee joint since the ankle is most likely unable to assist in absorbing landing forces in this landing technique. It is plausible to suggest that the participants, while performing a rearfoot landing technique, are placing greater demands on the knee joint by using a heel-to-toe strategy and minimizing the force absorption by the ankle joint and calf musculature. This is further supported by Boden and colleagues that conducted a 2-D video analysis of foot position at time of injury and reported that at the time of injury the basketball athletes were primarily in a dorsiflexion position at ground contact (Boden, et al., 2009). The International Olympic Committee has recently recommended a forefoot landing strategy as a protective mechanism for ACL injuries (Renstrom, et al., 2008). The authors speculate that this is a defensive mechanism of the knee joint based on results and observations of various neuromuscular intervention programs. Lebiedowska and colleagues using a simulation approach of various landing techniques (heel, toe, and mid-foot) reported that foot-landing position creates different characteristics of body stiffness and damping. The heel landing presented the highest stiffness, whereas a toe landing decreased body stiffness in half. (Lebiedowska, Wente, & Dufour, 2009) Similarly to experimental studies, it was argued that the decrease in body stiffness during the toe landing might be a result of the ankle dorsiflexion action, which is not observable during the heel-toe landing. The stiffness generated by a heel landing is most likely directly transmitted to the knee joint. This in turn is likely to increase the strain experienced by the knee ligaments, including the anterior cruciate ligament.

It has been noticeable in the literature that the tasks used to evaluate those risk factors have diverged, with drop-jump, stop-jump, sidestep cutting, and pivot tasks being utilized across different experiments. The drop-jump task has been one of the most commonly used tasks to evaluate participants landing patterns (Baca, 1999; Bobbert, et al., 1987a, 1987b; Bobbert, Mackay, Schinkelshoek, Huijing, & van Ingen Schenau, 1986; Cortes, et al., 2007a; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Ford, et al., 2003; Kovacs, et al., 1999; Russell, et al., 2006; Salci, et al., 2004b; Self & Paine, 2001). A few studies have used a running stop task (Chappell, et al., 2007; Chappell, et al., 2005; Chappell, et al., 2002b; Ford, et al., 2005; Yu, et al., 2006; Yu, et al., 2004; Yu, et al., 2005). The sidestep cutting task has been used to mimic a deceleration and cutting motion similar to the hypothesized risk mechanism (Cowley, et al., 2006; Dempsey, et al., 2007; Houck, 2003; Houck & Yack, 2003; Houck, Duncan, & Haven, 2006; McLean, Huang, et al., 2004; McLean, Lipfert, & van den Bogert, 2004; McLean, Myers, Neal, & Walters, 1998; McLean, et al., 1999; Pollard, Davis, et al., 2004b; Pollard, Heiderscheit, et al., 2004; Pollard, et al., 2007; Sigward & Powers, 2007).

The drop-jump task permits a controlled environment to serve as a baseline measure prior to applying more dynamic/sports related tasks; the control of this task can be comparable with some sport situations (i.e., basketball rebound). The sidestep cutting task attempts to replicate a real-life task to the laboratory environment, for example cutting in a soccer game (McLean, Huang, et al., 2004; Pollard, et al., 2007). There are some natural differences between these tasks. The drop-jump is a simple drop from a box with no change of direction involved, whereas sidestep cutting requires a deceleration and acceleration phase while simultaneously performing a cutting motion. A cutting action that includes a deceleration with a rapid change in direction has been related to the mechanism of injury (McLean, Lipfert, et al., 2004). Greig (2009) argued that the sidestep cutting does not replicate the demands of a pivot task that normally occurs during a soccer game (Greig, 2009). A pivot task, with 180 degrees of change in direction, was reported to provide a more realistic representation of a soccer task (Greig, 2009). This 180-degree maneuver commonly seen in soccer requires a complete deceleration with a change in direction followed by acceleration to maximum speed. These inherent differences suggest that the control mechanism and demands between these tasks are naturally different, and the multiple biomechanical risk factors may have a dissimilar role depending on the task, since it may require different demands from the motor system (Newell & Slifkin, 1998a; Newell & Corcos, 1993b).

Few studies have attempted to quantify and compare biomechanical parameters among tasks. The understanding of how the hypothesized risk factors behave under different task constraints might provide better insight into augmented risk motions. The problem with the intrinsic difference in the control mechanisms of various tasks, combined with the factor of how those tasks are conducted under laboratory experiments has been of recent concern. Researchers have focused on creating a more realistic approach through the use of light stimulus to produce an unanticipated factor (Beaulieu, et al., 2008; Ford, et al., 2005; Pollard, Heiderscheit, et al., 2004). The light stimuli do not truly mimic a game situation, although an improvement over standard laboratory setting the environment that players normally experience is still not present under this situation. Consequently, it is essential that more realistic scenarios are developed and ultimately utilized when evaluating biomechanical parameters related to ACL risk factors. This approach to a real-life situation attempts to improve a study's ecological validity, which is often underestimated and undervalued (Robins, et al., 2008; Shiffman, et al., 2008). The applicability and generalization of any study to real-world situations is dependent on its design (Robins, et al., 2008; Shiffman, et al., 2008). Neuropsychologists have started to focus on this factor, and have investigated the effect of conducting the studies under real-life situations (Chaytor, et al., 2006; Chaytor, et al., 2007). Recently, Parsons and colleagues have implemented a virtual reality environment to study neurocognitive functions, which has shown to improve its reliability and (ecological) validity (Parsons, et al., 2008).

CHAPTER III

Experiment I – Differences in the Pattern of Coupling Relations between Drop-Jump and Sidestep Cutting Actions

Title: Differences in the Pattern of Coupling Relations between Drop-Jump and Sidestep Cutting Actions

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Introduction

Rather than being viewed simply as noise (Harris & Wolpert, 1998; Schmidt, et al., 1979), variability of motion is considered an inherent characteristic of the motor system and movement performance (Newell & Corcos, 1993b). Indeed, for many voluntary actions, the presence of increased variability can be beneficial to movement performance since it affords the individual the capacity to respond optimally to different task challenges, and subsequently reducing the likelihood of potential injuries (Hamill, et al., 1999; Holt, et al., 1996; Neuringer, 2002; Neuringer, 2004; Newell & Corcos, 1993a; Newell & Slifkin, 1998a; Yates, 1987). The use of variability measures to assess movement performance has been shown to be particularly useful in a variety of contexts. For example, changes in the variability of motion can discriminate between individuals on the basis of injury (Hamill, et al., 1999; Heiderscheit, 2000; Heiderscheit, et al., 2002b), gender (Barrett, et al., 2008), neurological disorders (Dingwell & Cusumano, 2000; Dingwell, et al., 1999; Hausdorff, et al., 1998; Hausdorff, et al., 1997), and normal ageing (Hausdorff, et al., 1996; Hausdorff, et al., 2001).

The assessment of changes in movement variability has proven to be particularly useful for assessing adaptation to or risk of injury (Hamill, et al., 1999; Heiderscheit, 2000; Heiderscheit, et al., 2002b; Pollard, Davis, et al., 2004a; Pollard, et al., 2007). Hamill and colleagues (1999) reported that individuals with unilateral patella-femoral pain exhibit low variability in joint coupling during rapid, cutting maneuvers. It was subsequently argued that, because of this loss of variability, these individuals may have a reduced ability to adjust to the task demands, an outcome which potentially places them at greater risk of injury (Hamill, et al., 1999). A similar result was reported by Pollard et al. (2005) where females exhibited lower variability in lower limb joint coupling during an unanticipated cutting maneuver. This diminished variability was argued to represent a risk for injury because of greater localized mechanical stress on anatomical structures that may contribute in the longer term to degenerative changes from overuse (Pollard, et al., 2005a). Both the studies by Hamill (1999) and Pollard (2005) where designed to assess the impact of different running tasks (cutting maneuvers) on lower limb injury. One common theme of this research is to identify those factors that could contribute to damage the anterior cruciate ligament (ACL), one of the most debilitating knee ligament injuries in the collegiate athletic population (Agel, et al., 2005; Arendt & Dick, 1995).

Despite the insights that changes in movement variability provide into injury assessment in lower limb activities, our understanding of the risk factors for ACL injury is confounded by the fact that the measured movement outputs vary depending on the level of the motor system at which the output is assessed (Newell & Corcos, 1993a; Newell & Slifkin, 1998a). For ACL injury, this is highlighted by the fact that numerous biomechanical factors across different joints and actions have been identified as potential markers for injury (Blackburn & Padua, 2008; Ford, et al., 2003; Houck, 2003; McLean, Huang, et al., 2004; Sell, et al., 2007; Yu & Garrett, 2007). For example, some of the variety of different joints/factors which have been theorized as potential risk factors include the knee (decreased knee flexion, increased knee valgus), the hip (decreased hip flexion), the tibia (increased proximal anterior tibia shear force) and the foot/leg as a whole (decreased peak vertical ground reaction force) (Blackburn & Padua, 2008; Ford, et al., 2003; Houck, 2003; McLean, Huang, et al., 2004; Sell, et al., 2007; Yu & Garrett, 2007).

The problem of clearly identifying biomechanical risk factors for ACL injury is further compounded by the fact these risk factors change as a function of the specific task being performed (Newell & Corcos, 1993a; Newell & Slifkin, 1998a). In this regard, the movement output reflects and is affected by the specific parameters of the action itself (task dependent factors). For ACL injury, the predictive variables that can be assessed alter as a function of the movement being performed, that is, whether the resultant action involves horizontal deceleration, vertical deceleration, and/or rotation. For many predictive injury studies, two common movements have been used to assess ACL risk factors; namely sidestep cutting (McLean, Huang, et al., 2004; McLean, et al., 2008; Pollard, et al., 2007; Sigward & Powers, 2006a), and the drop-jump (Chappell & Limpisvasti, 2008; Cortes, et al., 2007a; Ford, et al., 2003; Kernozek, et al., 2005). Arguably, the sidestepping task has been more commonly employed because of its close association with real-life athletic tasks (McLean, Huang, et al., 2004; McLean, et al., 2008; Pollard, et al., 2007; Powers, et al., 2004). The drop-jump task has been utilized primarily because its landing control is comparable with other athletic tasks (e.g., sidestep cutting task), and from the experimenters' perspective it is easier to perform under controlled laboratory settings (Noyes, et al., 2005; Yu & Garrett, 2007). The drop jump task has also been utilized since several ACL injuries have been linked with a vertical drop landing from a jump, such as a rebound in basketball.

However, there are some inherent differences in the movements themselves. As the name implies, the drop-jump entails a vertical drop from a box with minimal-to-no rotational component of the lower limb segments. Conversely, sidestep cutting contains a significant horizontal velocity and rotational component (i.e., internal rotation of the knee) due to the change in direction (i.e., 45 degree angle), two features that are not present in the drop-jump task. Furthermore, the sidestep cutting includes a deceleration/acceleration phase – an important distinction since a rotational component is often associated with ACL tears. Given these intrinsic task differences, there is little wonder that a variety of different potential biomechanical variables across multiple lower extremity joints have been identified as risk factors. However, despite the numerous factors identified, their occurrence over multiple locations within the lower limb, and the task dependent nature of the injuries, most studies have focused on reporting single risk factors as the leading cause of ACL injury (McLean, Huang, et al., 2004; Yu & Garrett, 2007). What is apparent is that there are multiple factors which can contribute to ACL injury and that these factors probably alter as a function of the task and population being observed. In order to gain a clearer understanding of the mechanisms of ACL injury, it is essential to identify what the risk factors are, how the different factors are actively related or coupled and whether differences in these coupling relations can be observed across different tasks.

One concern is that most studies have focused on single predictive variables when comparing a drop jump task with sidestep. Typical single variables have included knee flexion, knee valgus, hip flexion, hip rotation, and/or ground reaction forces (Blackburn & Padua, 2008; Chappell & Limpisvasti, 2008; Cortes, et al., 2007a; McLean, et al., 2007; Pollard, Davis, et al., 2004a). Given the multitude of potential contributing risk factors and the high likelihood of strong interactions between factors, this approach is unlikely to provide a clear insight as to the mechanism of ACL injury. The primary aim of this project was to identify kinematic and/or kinetic variables that are descriptors of each movement using a principal component analysis method. A secondary aim was to compare the variability of selected kinematic and kinetic variables between the drop-jump and sidestep cutting task as measured by the coefficient of variation.

Methodology

Participants. Nineteen female collegiate soccer athletes (age = 19 ± 0.8 years; height = 1.67 ± 0.05 meters; mass = 63.7 ± 10.1 kg) from a Division I institution participated in this study. Females were selected since it has been suggested that they are typically at higher risk of injury than males (Agel, et al., 2005; Arendt & Dick, 1995; Tillman, et al., 2005). All participants were screened using a validated questionnaire prior to inclusion to ensure none had any previous hip, low-back, knee, or severe ankle injuries within the last six months or surgeries within the last 2 years (Onate, et al., 2005). Each participant performed the specified tasks with their dominant leg which was defined as the leg that the subject would use to kick a soccer ball as far as possible (Ford, et al., 2003; Hewett, Myer, & Ford, 2005). Prior to data collection, approval of the research through Institutional Review Board of Old Dominion University, and written informed consent form for all participants was obtained.

Experimental Procedure. For all testing procedures, clothing and footwear was standardized between subjects. All individuals wore spandex shorts, sports bra and the team running shoes (Adidas Supernova, AG, Herzogenaurach, Germany). General anthropometric measures were taken for each participant. This included weight, height, knee width, ankle width, elbow width, wrist width and hand thickness. The same researcher completed all anthropometric measures. Each person then completed a 10-

minute warm-up period of cycling and self-directed stretching. After the warm-up, thirty-five reflective markers were placed on specific body landmarks according to a modified Helen Hayes model (Kadaba, Ramakrishnan, & Wootten, 1990; Kadaba, et al., 1989). Before the testing period commenced, a static standing trial with the participants standing on the force plates with shoulders abducted at 90 degrees was obtained. The static trial (calibration) was used to compute the kinematic model and calculate the various biomechanical variables of interest.

All individuals were required to participate in two different movement tasks, a drop jump and a sidestep cutting. For the drop jump, the participants stood on a 30 cm box placed 30 cm from the force plates. They were instructed to shift their weight forward to initiate the movement by inclining their trunk forward, and then drop from the box onto the force plates to execute each trial. After landing on the force plates, participants were instructed to immediately jump as high as they could straight up in the air ("as if they were performing a soccer header"), and finally land back on the force plates. At that time the entire foot needed to be on the force plate, with each foot on a separate force plate. The initial landing from the box was used for the purpose of analyses with the secondary landing being discarded. Each participant performed three practice trials followed by three successful trials, with 1-minute rest period between trials to minimize the effects of fatigue.

For the running sidestep cutting, participants stood on the beginning of the running platform pressing a footpad to trigger the speed-timing device. Participants started running at their leisure once the timers were engaged and stepped with the dominant foot onto one of the force plates. At that moment they had to perform a cutting motion to the contra-lateral side of the dominant foot touching the force plate. A custom made platform placed with an angle between 35 to 55 degrees relative to the force plates, was used to achieve a cutting angle of approximately 45° during the sidestep task (McLean, Huang, et al., 2004). A Brower timing system (Brower Timing Systems, Draper UT, USA) was used to control the approach speed. Participants had an approach speed of 3.66 ± 0.26 $m.s^{-1}$ (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007). Participants were permitted three practice trials. Five successful trials were collected for each task. A 2-minute rest period was provided between trials to minimize fatigue.

Instrumentation. Kinematic measures of the body segments were attained using the VICON motion capture system with eight high-speed video cameras (MX-F40, Vicon Motion Systems Ltd., Oxford, England). Kinetic data relating to the ground reaction forces were attained from two Bertec force plates (Model 4060-NC, Bertec Corporation, Columbus OH, USA). The sampling rate for the cameras and the force plates was 500 Hz.

From the standing (static) trial, a full body kinematic model was created for each participant using Visual 3D (C-Motion, Rockville MD, USA). This kinematic model was used to quantify the motion at the hip, knee, and ankle joints. A Cardan angle sequence (x-y-z) was used to calculate joint angles, which is comparable to a joint coordinate system (Grood & Suntay, 1983). The pelvis was modeled as a cylinder and the lower extremity as frusta of cones. Based on a power spectrum analysis, all kinematic and kinetic data were low-pass filtered through a fourth-order Butterworth zero lag filter with a 25 Hz cutoff frequency.

Data Analysis. Principal Component Analysis (PCA): This analysis has been extensively used for tasks that are multi-factorial in nature, primarily to identify the variables that are highly associated with the data variance (Landry, et al., 2007; Muniz & Nadal, 2009). PCA determines a set of factors, or principal components, that describe the variation in the data. These factors generate a component matrix with loading coefficients for each variable. The correlation between the variables and each principal component is represented by coefficients (Berenson & Levine, 1983). The larger coefficient per variable, the higher its correlation with a component of interest. Each principal component is uncorrelated as it measures a different dimension within the data set, and assumes that the optimization of the results can be obtained if several original variables are highly correlated (Manly, 1988). The principal component analysis was employed to selected variables from the waveform.

The eigenvalue (EV) criterion was used to explain the total variance by each principal component. Any principal component with an eigenvalue greater than or equal to 1.0 and which also accounted for over 10% of the variation in the data was included (Berenson & Levine, 1983; Manly, 1988). This was used to establish the correlation between the variables and the principal components previously found. Those coefficients were orthogonally rotated (varimax) with Kaiser normalization, to maximize the variance of each coefficient. The correlation between each principal component and variables were calculated (Berenson & Levine, 1983). A positive or negative correlation of 0.6 was the criteria to include the variable (Manly, 1988).

Coefficient of Variation: The coefficient of variation (CV), measured as the standard deviation divided by the mean, and respective standard error of the mean (SEM) was used to quantify the variability of specific kinematic variables during each task. The CV was calculated for the entire stance phase of each dependent measured. The specific variables assessed were; ankle flexion, hip flexion, hip abduction, knee flexion, knee valgus, and trunk flexion.

Statistical Analysis. Prior to these analyses, case-wise diagnostics were performed to assess data normalcy. All statistical analyses were conducted using SPSS (version 16.0, SPSS Inc, Chicago IL). The kinematic and kinetic data were analysed using principal component analysis and coefficient of variation. Separate paired *t*-tests were conducted to assess significant differences in the CV between the two tasks for each dependent measure. The alpha level for statistical significance was set *a priori* at $p < 0.05$.

Results

An example of the typical pattern of ankle, knee, and hip flexion angles for the drop jump and sidestep cutting are shown in Figure 1 (Appendix I). The solid area represents the standard deviation for all trials during the stance phase of the drop-jump, while the dashed area represents the standard deviation for all trials during the stance phase of the sidestep cutting.

Principal Component Analysis. While the results of the PCA revealed a number of components for both the drop-jump and sidestep tasks had eigenvalues above 1, most accounted for only a small proportion (< 9%) of the variation in the data. Only three components for each task were found to have eigenvalues above 1 and account for at least 10% of data variance. These three factors were included for further analysis. Tables 1 and 2 contain the variables which made up the first and second components for both the drop jump and the cutting maneuver, respectively (Appendix II & III). Each table contains the correlation values for each variable, the eigenvalues and the total variance explained by each component.

Drop Jump Task. For this task, the three factors identified by the PCA contained only kinematic variables and together these groupings accounted for 51% of the variance in the data. The first principal component explained 23% of the variance and included kinematic parameters related to trunk flexion and knee valgus. The second component included the kinematic variables of ankle flexion, knee valgus, and hip rotation. This factor explained 15% of the data variance. The third component included variables of the knee and hip joints (e.g., hip flexion, hip abduction and rotation, and knee flexion and valgus). This component explained 13% of the variance.

Sidestep Cutting Task. The three variable groupings identified for this task accounted for 47% of the variance in the data. Unlike the drop jump, the variables identified within this task included both kinematic and kinetic measures. The first principal component, which explained 18% of the variance, included those kinematic variables primarily related with the ankle joint, while the knee and hip joints accounted for the kinetic aspects (e.g., proximal anterior tibia shear force and hip flexion moment). The second component explained 16% of the data variance but only included kinematic variables. These were those measures related to ankle flexion, knee flexion, and hip flexion. The third component only included the knee valgus motion at different time instants (e.g., initial contact, maximum knee flexion). This component explained 12% of the variance.

Coefficient of Variation and Standard Error of the Mean. Principal components 1 and 2 combined explained 38% and 34% of the data variance during the drop jump and sidestep cutting task. Coefficients of variation were calculated for each variable that highly correlated with either one of the maneuvers. The results of this analysis demonstrated a significant difference in the coefficient of variation for the ankle flexion, hip flexion, and knee flexion between the drop jump and sidestep cutting tasks ($t_{2,82}$=4.35, P<0.05 (Figure 2, Appendix IV). The sidestep cutting task always had significantly higher variation (higher CV) than the drop jump for ankle, hip, and knee flexion. No significant difference between the two tasks was observed for any of the remaining measures (P>0.05) (Figure 3, Appendix V).

Discussion

The current study examined the pattern of lower limb coupling relations for a drop-jump and a side-step cutting actions. These two movements were selected since they are used as standard laboratory tasks to evaluate biomechanical risk factors for ACL injury. Principal component analysis was employed in order to assess the relations between those factors that make up these actions. These results demonstrated that the highly loaded biomechanical measures varied across the two movements, showing that the factors are inherently different depending on vertical versus horizontal oriented jump-landing tasks. Specifically, trunk flexion and knee valgus were highly correlated with the first component for the drop jump. Whereas for sidestep cutting, the variables that highly correlated within the first component were ankle and hip flexion, and knee proximal anterior tibia shear force. In addition, the sidestep cutting had higher coefficients of variation for knee, hip, and ankle flexion than observed during the drop-jump. This differentiation demonstrates that the biomechanical movement patterns are different for each task and that the decreased variability during the drop jump may result in different adaptability of the system.

Task Differences in the Pattern of Joint Coupling. Principal component analysis has been shown to be a valuable tool in identifying different movement relations in a number of different contexts including gait analysis (Lee, Roan, Smith, & Lockhart, 2009; Rugelj, 2009), postural sway (Oliveira, Simpson, & Nadal, 1996), balance control and falls risk (Rugelj, 2009), upper limb ballistic actions (Morrison & Anson, 1999), and in the orthopaedic biomechanics field (Landry, et al., 2007; Muniz & Nadal, 2009). Given the multitude of variables in drop-jump and sidestep cutting, this analysis also has the potential to provide insight into the kinematic/kinetic coupling relations during these movements (Landry, et al., 2007). The main advantage of PCA for evaluating biomechanical variables during these two selected tasks over more classical approaches is instead of evaluating a single time point, it evaluates the variable over the entire time series (Muniz & Nadal, 2009).

In the context of our current study, the PCA highlighted which joints and motions have stronger coupling relationships within each maneuver – a result which can be relevant to distinguish the descriptors of each task. Despite the previous trend to only assess and identify single factors (McLean, Huang, et al., 2004; Yu & Garrett, 2007; Yu, et al., 2005), the current findings demonstrated that there was not a single joint and/or motion highly correlated with each movement, but rather multiple factors were highly related. This finding indicates that ACL injuries related to either of these tasks are probably not due to a single, isolated mechanism, but are rather multi-factorial. Consequently, when these factors are combined it may increase the risk of ACL injury. Given that the prevalent view that one primary mechanism is the main cause of injury during tasks of this nature (McLean, et al., 2008; Yu, et al., 2005), this result is of some clinical significance. The results of the PCA analysis demonstrated that for the drop-jump, trunk flexion, hip abduction and rotation, knee valgus, and ankle flexion explain approximately 40% of the data variance. Whereas, for sidestep cutting, hip flexion, hip flexion moment, proximal anterior tibia shear force, knee flexion, and ankle flexion account for a similar proportion (34%) of the variance in the data. While factors such as small knee flexion (Blackburn & Padua, 2008; Yu & Garrett, 2007; Yu, et al., 2005), and excessive proximal anterior tibia shear force (Sell, et al., 2006, 2007) have been proposed as principal causes for ACL rupture, this has been contrasted by other studies which have reported that sagittal plane motion cannot tear the ACL (Hewett, Myer, & Ford, 2005; McLean, Huang, et al., 2004). Indeed, Hewett and colleagues have proposed that knee valgus/abduction angles and moments are the primary injury mechanism (Ford, et al., 2003; Ford, et al., 2005; Hewett, Myer, & Ford, 2005; Sigward & Powers, 2007). More recently some authors have even suggested that hip rotation might be the main leading reason to injury (Pollard, Davis, et al., 2004a; Pollard, et al., 2007).

Task Specific Changes in Variability. Many authors have argued that variability, rather than being seen as random fluctuations or noise (Harris & Wolpert, 1998), is instead a highly functional characteristic of the human motor system (Bassingthwaighte, Liebovitch, & West, 1994; Davids, Bennett, & Newell, 2006; Newell & Slifkin, 1998a; Newell, Challis, & Morrison, 2000; Yates, 1987). The results of the coefficient of variation analysis demonstrated significant differences between the drop jump and sidestep cutting tasks for ankle, knee, and hip flexion. In particular, increased variability in these kinematic parameters was observed during the sidestep cutting maneuver. The differences between the two tasks highlights that, in spite of the perceived similarities between these two actions (Cortes, et al., 2007a; Ford, et al., 2003; Kernozek, et al., 2005; McLean, Huang, et al., 2004; McLean, et al., 2008; Pollard, et al., 2007; Sigward & Powers, 2006a), the dynamics of each movement are inherently different. Not only do these tasks differ with regard to the variables which describe each action, the degree of variability for the same measures differs significantly between the two movements. The higher degree of variability observed for the ankle, knee and hip motion for the side-step may demonstrate the greater potential for modifying the movement response. Contrastingly, the performance of the drop jump would appear to be more restrictive in terms of the degree of variability that any given individual can exhibit. While increased variability is theorized to reflect enhanced adaptive control and greater flexibility of performance (e.g., the individual is able to respond optimally to different task challenges) (Newell & Slifkin, 1998a; Newell, Vaillancourt, & Sosnoff, 2006), in this situation, it appears that the task constraints inherent within the drop jump activity imposes a greater level of external restriction on the movement dynamics in comparison to the sidestep action.

Some authors have reported that low variability, especially in joint coupling, is related to the lack of the participants' ability to adjust to the task demands, and consequently place them at greater risk of injury (Hamill, et al., 1999). The increased variability may be a protective mechanism under unanticipated situations where there is a need to quickly adjust to several stimuli (e.g., opponent player, changing direction). The decreased variability obtained on the drop-jump does not necessarily represent an increased risk of injury, but rather the consequence of a highly controlled environment.

Differences in the pattern of coupling were also observed. The results demonstrated changes in the coupling relations at the kinematic level, but less so in regards to the kinetics of the respective actions. This finding supports the view that differences in the variable nature of the movement output may not be observed at all levels of the motor system (Newell & Slifkin, 1998a). At the joint level, our findings highlighted that each movement was characterized by coupling relations than spanned multiple lower limb locations and were not localized to a single area or structure. The observed differences in the pattern of coupling across both movements is consistent with the general findings of van Emmerik et al. (1999) and Barrett and colleagues (2008) who similarly showed that variability assessed at the joint level is more likely to provide insight as to specific group and/or task differences (Barrett, et al., 2008; Van Emmerik, et al., 1999).

Kinematic and Kinetic Contributions to Task Performance. Within the sports medicine literature, it has been proposed that the mechanisms of ACL injury are similar for both sidestep cutting and drop jump because of the similar nature of the landing control (Blackburn & Padua, 2008; Noyes, et al., 2005). However, the findings of the current study revealed that specific differences in the kinematic and kinetic descriptors existed between these two actions. As previously highlighted, the PCA results demonstrated that the critical variables explaining the variance were not the same for each movement. However, what also emerged is that the kinematics of the lower extremity has a higher correlation with each task than the forces produced. For example, while kinematic variables were shown to be primarily correlated with each movement (i.e., hip flexion, knee flexion, knee valgus, hip rotation), only a few kinetic variables emerged (i.e., proximal anterior tibia shear force, knee valgus/abduction moment). The relevance of the differences found between jump-landing actions is important within the context of injury prevention since it illustrates that jump-landing biomechanical variable contributions are different for these two actions. A result that is contradictory to the aforementioned perspective that these tasks can be viewed as similar with regard to the risk of ACL injury (Blackburn & Padua, 2008; Noyes, et al., 2005).

Conclusions

Overall, kinematic variables were shown to have a stronger coupling relationship with both tasks. In addition, the drop jump was also characterized by different variables as compared to the sidestep cutting. This cutting task displayed increased variability in hip, knee, and ankle flexion. All of these factors combined indicates that the mechanism of injury is possibly different between a vertically oriented drop jump as compared to a sidestep cutting that requires rapid horizontal oriented deceleration/acceleration combined with change of direction. The increased variability in the sidestep task can be seen as a protective mechanism due to the adaptability of the participants' to the increased challenge demands of the sidestep task.

CHAPTER IV

Experiment II – Pivot Task Increases Knee Frontal Plane Loading When Compared to Sidestep and Drop-Jump

Title: Pivot Task Increases Knee Frontal Plane Loading When Compared to Sidestep and Drop-Jump

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Introduction

Extensive research efforts have been applied to understand the mechanism of anterior cruciate ligament (ACL) injury (McLean, Huang, et al., 2004; Yu, et al., 2006). The devastating consequences of ACL tears (health and monetary) and undefined consensus regarding the mechanism of injury have placed it as a major topic of biomechanical research. Numerous risk factors have been conjectured as probable mechanisms of injury (Davis, Ireland, & Hanaki, 2007; Griffin, et al., 2006). These risk factors have received much attention from researchers with prevention programs being implemented in attempts to modify them; however, the rate of injury has remained steady over the past decade (Agel, et al., 2005; Prodromos, Han, Rogowski, Joyce, & Shi, 2007). A multitude of biomechanical risk factors across the lower extremity joints have been hypothesized as potential markers for injury; ranging from decreased knee flexion angle at initial contact, increased knee valgus angle and loading, high peak vertical ground reaction force, and increased proximal anterior tibia shear force (McLean, Huang, et al., 2004; Sell, et al., 2006, 2007; Yu, et al., 2006). Some authors have supported sagittal plane kinematics and kinetics as the primary risk factor, while others claim that the sagittal plane biomechanics cannot injure the ACL, arguing that knee valgus angle and valgus loading are the primary risk factors for injury (McLean, Huang, et al., 2004; Yu & Garrett, 2007; Yu, et al., 2006). One consideration that should be noted is the comparison of methodological approaches between these studies. The tasks used to evaluate the relation of risk factors and increased likelihood of injury has differed slightly, yet investigation of the effects of task on movement performance is limited. In McLean's (2008) study, they analyzed a sidestep cutting task; whereas Yu and colleagues

(2006) evaluated a running stop task. Concepts of motor learning indicate that any outcome is directly influenced by the triangle of task (i.e., movement dynamics to achieve the goal), person (e.g. unique characteristics of organism/individual), and environment (e.g., terrain and noise) (Newell, 1996). The organization of the individual to successfully accomplish the movement task goal could possibly explain the different results. It is important to understand the effects various jump-landing movement tasks have on lower extremity biomechanics.

It has been noticeable in the literature that the tasks used to evaluate biomechanical risk factors have diverged across experimental protocols with various types of tasks (e.g., drop-jump, stop-jump, sidestep cutting, and pivot) utilized across different experiments, yet discussed as having similar types of movement demands. The drop-jump task has been commonly used to evaluate participants' landing patterns (Cortes, et al., 2007b; Kernozek, et al., 2005). This task permits a controlled environment to serve as a baseline measure prior to utilizing dynamic/sports related tasks. The sidestep cutting task (McLean, Huang, et al., 2004; Pollard, et al., 2007; Sigward & Powers, 2006a) has been used to mimic a deceleration and cutting motion similar to the hypothesized risk mechanism. It also attempts to replicate a real-life task in a laboratory environment (i.e., cutting in a soccer game) (McLean, Huang, et al., 2004; Pollard, et al., 2007). Other researchers have focused on a running stop task (Chappell, et al., 2002b; Yu, et al., 2006), which requires a sudden deceleration and a jump straight into the air to reproduce a soccer header. There are some natural differences between these tasks. The drop-jump is a simple drop from a box with no dynamic motion nor decision process involved (anticipated), whereas sidestep cutting requires a deceleration and acceleration phase while simultaneously performing a cutting motion. The deceleration-acceleration phase with rapid change in direction has been observed during ACL injury events (McLean, Lipfert, et al., 2004). Greig (2009) argued that the sidestep cutting does not replicate the demands of a pivot task that normally occurs during a soccer game. A pivot task, with 180 degrees of change in direction, was reported to provide a more realistic representation of a soccer task (Greig, 2009). This maneuver requires a complete deceleration with a change in direction followed by acceleration to maximum speed. These inherent differences suggest that the control mechanism and demands between these tasks are different, and the multiple biomechanical risk factors may have a different role depending on the task (Newell & Slifkin, 1998a; Newell, 1996). Few studies have attempted to quantify and compare biomechanical parameters among tasks. The understanding of how the hypothesized risk factors behave under different task constraints might provide better insight into possible risk motions. The intrinsic difference in the control mechanisms of various tasks and how those tasks are conducted under laboratory experiments has been of recent concern. The purpose of this study was to determine kinematic and kinetic differences between three tasks (drop-jump, sidestep cutting, and pivot tasks) commonly associated with anterior cruciate ligament injuries. We hypothesized that drop-jump tasks would produce higher knee and hip flexion angles, decreased knee valgus angles and loading, and vertical and posterior ground reaction forces when compared with two unanticipated tasks (pivot and sidestep cutting).

Methodology

Participants. An *a priori* power calculation was conducted to estimate the sample needed to establish differences between athletic tasks. Using data from the literature (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Ford, et al., 2006; Lephart, et al., 2002a), for a power level of 80% and an alpha level of 0.05, a necessary sample size ranged from 14 to 20 participants. Nineteen female collegiate soccer athletes (age = 19.6 ± 0.8 years old; height = 167 ± 5 cm; mass = 63.7 ± 10.1 kg) from a Division I institution were chosen to participate in this study. Prior to data collection, approval of the research through Institutional Review Board, and written informed consent for all participants was obtained. Participants were screened to ensure none had any previous hip, low back, knee, or severe ankle injuries within the last six months or surgeries within the last 2 years. The dominant leg, defined as the leg that the participant would use to kick a soccer ball as far as possible, was used for analysis.

Experimental Procedure. Participants wore spandex shorts, sports bra and the team running shoes (Adidas Supernova, AG, Herzogenaurach, Germany). Participants completed a 5-minute cycling warm-up and 5-minute self-directed stretching. General anthropometric measures were taken for each participant. Reflective markers were placed on specific body landmarks according to a modified Helen Hayes marker set (Kadaba, et al., 1990). A standing (static) trial with the participants standing on the force plates with shoulders abducted at 90 degrees was obtained. The static trial was later used to compute the kinematic model.

Participants were required to conduct three movement tasks; drop-jump, sidestep cutting, and pivoting maneuver. The drop-jump was performed upfront with the other two tasks being randomly generated. For the drop-jump task, the participants stood on a 30 cm box placed 30 cm from the force plates. They shifted their weight forward to initiate the movement by inclining their trunk, and dropped from the box onto the force plates as vertically as possible. After landing, participants were instructed to immediately jump as high as they could straight up in the air, and land back on the force plates. The initial landing from the box was used for the purpose of analyses with the secondary landing being discarded. Each participant performed a total of three successful trials, with 1-minute rest period between trials to minimize the effects of fatigue.

A custom-made visualization software was developed to randomly generate the sidestep and pivot tasks by creating an unanticipated event. It allowed the participants to see a soccer field, a soccer ball, and players projected onto a screen (Figure 1). The cues to either perform a sidestep cutting or a pivot task were based on the virtual player position. If a virtual player would show on the right side of the screen, the participants had to perform a sidestep cutting task. However, if the virtual player would show up in the middle of the screen, the participants had to plant with the dominant foot and pivot 180 degrees. Two meters from the force plates, an infrared beam was placed across the platform where the participants ran. When the participants crossed the infrared beam it triggered the software program to randomly generate the athletic tasks. The unanticipated factor was expected to mimic as as possible a soccer game situation, and provide stronger ecological validity to the experiment. A Brower timing system (Brower Timing Systems, Draper UT, USA) was used to control the approach speed. For the sidestep cutting task, participants ran and stepped with the dominant foot on the force plate. At that moment they had to perform a cutting motion to the contra-lateral side of the dominant foot touching the force plate. The running pathway was constrained to 35 to 55 degree angle to provide an optimal cutting angle of 45-degrees (Colby, et al., 2000). For the pivot task, participants ran and planted onto the force plate with the dominant foot, pivoted 180 degrees, and ran back to the starting position (Greig, 2009). Participants were permitted three practice trials, and then five successful trials were randomly collected for each task. If participants did not plant on the force plate with the dominant foot, lost balance, or did not perform the appropriate task based on the cue generated by the software, the trial was not deemed successful and discarded from analysis. There was a 1-minute rest period between trials to minimize fatigue. Participants had an approach speed of 3.7 ± 0.3 m.s^{-1} for the sidestep cutting task, and 3.9 $\pm$ .5 m.s^{-1} for the pivot task.

Instrumentation. Kinematic measures of the various body segments were attained using eight high-speed video cameras (Vicon Motion Systems Ltd., Oxford, England). Kinetic data relating to the ground reaction forces were acquired from two Bertec Force Plates, Model 4060-NC (Bertec Corporation, Columbus OH, USA). The sampling rate for the cameras and force plates was set at 500 Hz. Single-leg analysis was used for kinematic and ground reaction force data. From the standing trial a lower extremity kinematic model was created for each participant, which included the pelvis, thigh, shank, and foot, using Visual 3D software (C-Motion, Inc, Germantown MD, USA). This kinematic model was used to quantify the motion at the hip, knee, and ankle joints. A Cardan angle sequence was used to calculate joint angles (Grood & Suntay, 1983). An optimal 7 Hz cutoff frequency was determined for raw trajectory marker data and 25 Hz cutoff frequency for ground reaction force data. A standard inverse dynamics analysis was employed to the trajectory marker data and ground force data to calculate joint moments (Winter, 2005). Segment inertial characteristics were estimated for each participant (Dempster, 1955). Intersegmental joint moments are defined as internal moments. As an example, a knee internal extension moment will resist a flexion load applied to the knee.

Data Analysis. All data were reduced using Matlab 6.1 (The MathWorks, Inc, Natick MA, USA) software with the creation of a custom-made program to export the variables of interest into a Microsoft Excel spreadsheet. The five trials were averaged and exported into SPSS version 16.0 (SPSS Inc, Chicago IL, USA) for data analysis. Repeated measure analyses of variance (ANOVA) were conducted to evaluate the kinematic (hip flexion, knee flexion, knee valgus, and ankle flexion) and kinetic (vertical and posterior ground reaction forces, knee extension-flexion and varus-valgus moment) parameters at different time instants (initial contact, peak vertical ground reaction force, and peak stance phase). The alpha level was set *a priori* at 0.05.

Results

Kinematics. Descriptive statistics with mean, standard deviations, and 95% confidence intervals are presented in Table 1. While performing the pivot task, participants had lower knee flexion ($F_{2,36}$=43.447, p<0.001) and increased knee valgus ($F_{2,36}$=34.681, p<0.001) at initial contact than for the drop-jump and sidestep cutting (p<0.05). A typical pattern of knee flexion and valgus are presented in figure 2 and 3. There was no difference among tasks for ankle flexion and hip flexion at initial contact (p>0.05). At maximum vertical ground reaction force, the pivot task had lower knee flexion ($F_{2,36}$=21.508, p<0.001), and increased valgus angle ($F_{2,36}$=22.175, p<0.001) than the drop-jump and sidestep (p<0.05). Furthermore, sidestep was also significantly different than the drop-jump, where the participants were in a varus position (p>0.05). The hip flexion ($F_{2,36}$=41.587, p<0.001) at maximum vertical ground reaction was higher for the drop-jump than the sidestep and pivot (p<0.05). The sidestep was also higher than the pivot (p<0.05).

For knee flexion at peak stance, participants went into higher flexion ($F_{2,36}$=235.283, p<0.001) on the drop-jump than the sidestep and pivot (p<0.05). Knee valgus at peak stance, the pivot task presented higher valgus angles ($F_{2,36}$=9.235, p<0.001) than the sidestep and drop-jump (p<0.05). Lastly, hip flexion peak stance was lower ($F_{2,36}$=52.770, p<0.001) in the sidestep than the drop-jump and pivot, as well as lower in the pivot than in the drop-jump (p<0.05).

Kinetics. Descriptive statistics are depicted in Table 2. At initial contact, the drop-jump had higher posterior ground reaction ($F_{2,36}$=12.864, p<0.001) than the sidestep and pivot tasks (p<0.05); however, at peak posterior ground reaction force the pivot had higher posterior ground reaction force ($F_{2,36}$=52.860, p<0.001) than the drop-jump and sidestep cutting (p<0.05). Participants had greater vertical ground reaction forces ($F_{2,36}$=6.525, p<0.001) at its peak on the sidestep cutting than on the pivot and drop-jump (p<0.05). The pivot task presented lower knee extension-flexion moment ($F_{2,36}$=66.671, p<0.001), and higher knee varus-valgus moment ($F_{2,36}$=30.667, p<0.001) than the drop-jump and sidestep tasks at initial contact. For knee extension moment peak stance, sidestep was higher ($F_{2,36}$=33.245, p<0.001) than the drop-jump and pivot (p<0.05); the drop-jump was also higher than the pivot (p<0.05). Typical pattern of knee varus-valgus moment and posterior ground reaction are represented in figure 4 and 5. The participants had higher knee varus-valgus moment ($F_{2,36}=26.768$, $p<0.001$) at peak stance for the pivot than the drop-jump and sidestep ($p<0.05$) with the sidestep higher than the drop-jump ($p<0.05$).

Discussion

The present study was designed to evaluate kinematic and kinetic differences among three landing tasks in a female collegiate soccer population using innovative visualization software. One of the main results to emerge from this study is that the three tasks appear to have distinct kinematic and kinetic characteristics; specifically, increased knee valgus position and loading, increased peak posterior ground reaction force, and decreased knee flexion angle for the pivot task when compared with the drop-jump and sidestep cutting tasks. The delineation between tasks may suggest that they have differentiated characteristics and that the injury mechanism may be task dependent, possibly requiring individualized prevention programs and screening processes.

We found that the pivot task presented significantly higher knee valgus angle and loading at initial contact and at peak stance when compared to the other tasks. For the drop-jump task, our results are in disagreement with those from Blackburn and Padua (2008). They reported a knee valgus angle of 6 degrees, whereas our participants were almost in a neutral alignment (0.8 degrees). This suggests that even with low knee flexion angle at initial contact, the participants were able to maintain their alignment without displacing the knee into a valgus position, which has been hypothesized as a risk factor (Hewett, Myer, Ford, et al., 2005; McLean, Huang, et al., 2005). This difference may be explained by the participants' background. Blackburn and Padua opted for recreational athletes, whereas our volunteers were division I collegiate soccer athletes who are trained to perform these tasks on a regular basis. Contrastingly, for the sidestep and pivot tasks the participants were always in a knee valgus position, which is comparable to the results from Ford and colleagues (Ford, et al., 2005). The valgus alignment might increase the load on the ACL, especially in the pivot task where they attained approximately 11 degrees of valgus position. This increase in knee valgus angle might be due to the demands of the task. The participants had to come to a full deceleration and perform a 180-degree change in direction, which entails a full rotation over the dominant foot. The drop-jump movement on the other had consisted of deceleration followed by a jump into the air without an unanticipated factor. Lastly, the sidestep task has a momentary deceleration and change of direction as opposed to a complete stop of forward momentum, thus lending itself to varying movement pattern results. The various movement goals per jump-landing imposes changes on biomechanical movement pattern outcomes that need to be taken into effect when comparing across tasks and biomechanical risk factors for ACL injury.

We hypothesized that a dynamic task (sidestep and pivot) would increase the knee internal varus moment when compared with the drop-jump. This was supported by our results, showing a large increase in the pivot task over the two other tasks. Multi-directional jump-landing tasks more closely replicate field situations, thus indicating a greater risk for injury due to the increased frontal plane demands as compared to the uni-directional drop-jump task. Researchers have promoted that dynamic valgus load may be the primary risk factor for the rupture of the ACL (Ford, et al., 2003; McLean, Huang, et al., 2004). This valgus position combined with an increased internal varus moment has been shown to increase the load placed at the ligament (Bendjaballah, et al., 1997). A prospective study by Hewett and colleagues (2005) found that knee valgus angles and loading were strong predictors for athletes that injured their ACL (Hewett, Myer, Ford, et al., 2005). The authors theorized that if a valgus angle and loading are present during a landing, it could place excessive strain on the ACL and rupture it. We have observed that the pivot task presented higher knee valgus angle and loading, which may represent an increased strain on the ACL during the execution of this common task in soccer. However, this pattern was not fully observed for the other tasks. This can suggest that the pivot task augments the load on the ligament and increases the likelihood for injury. When comparing our sidestep results to those provided by McLean et al. (2005) it is interesting to observe that the knee varus moments for our female participants (0.49 Nm/Kgm) are similar to those presented by their male athletes (0.45 Nm/Kgm) (McLean, Huang, et al., 2005). Although we did not have a male population, this may suggest that our females were at decreased risk of injury during a sidestep cutting as McLean et al. (2005) theorized for their male population. Lastly, while performing the drop-jump task the participants were always in a varus position with minimal knee varus loading. This can potentially indicate that the drop-jump does not elicit similar factors for knee loading and likelihood of injury as the pivot and sidestep tasks.

A second result to emerge from our results is the decreased knee flexion angle at initial contact and peak stance for the pivot task compared to the drop-jump. A decreased knee flexion angle at initial contact has been proposed as a risk mechanism for ACL tear (Wojtys, Ashton-Miller, & Huston, 2002). With low knee angles (0 to 30 degrees) the quadriceps muscles can place enough strain on the ACL to rupture it (Nisell, 1985). In our study, the athletes presented decreased knee flexion while performing a pivot task (24 degrees). The low knee flexion angle presented at initial contact might place the participants at higher ACL strain due to the combination of an erect posture with a probable increase in quadriceps activation (Blackburn & Padua, 2009). At peak stance, the participants were in slightly lower knee flexion angle for the pivot and sidestep cutting tasks than the drop-jump. We found that peak posterior ground reaction force was higher in the pivot task than the other two tasks. Researchers have shown a high correlation between posterior ground reaction force and proximal anterior tibia shear force (Sell, et al., 2007; Yu, et al., 2006). Proximal anterior tibia shear force is thought to create an anterior displacement of the tibia, thus increasing the strain on the ACL (Sell, et al., 2007; Yu, et al., 2006). When this force is too high, it is speculated it can lead to ACL rupture. Consequently, a high posterior ground reaction force for the pivot task may suggest that there is an increased load on the ACL. It is our perspective that the association of two theoretical risk factors, low knee flexion and increased posterior ground reaction force, is most likely increasing the knee loading and the demands in the knee ligamentous structures during the pivot task. It is worth noting that the participants experienced approximately one time their bodyweight for posterior ground reaction force during this task. However, caution is required when speculating concerning the link of a single variable to potential ACL tear since none of the participants actually injured their ACL during testing.

An interesting result to note is the decreased hip flexion range of motion during the sidestep and pivot tasks. The hip flexion range of motion, similarly to the knee flexion, was significantly higher in the drop-jump than in the sidestep and pivot. The augmented knee and hip range of motion for the drop-jump makes it plausible to presume that the athletes assumed a more protective landing mechanic towards the ACL during the drop-jump task than for the sidestep and pivot. However, this fact is possibly due to the nature of the tasks. The drop-jump motion can be observed while landing from a basketball rebound or landing from a soccer header, whereas the athlete have to quickly react and adjust to stimuli for the sidestep and pivot cutting motions (i.e., players, ball). Further investigation on the amount of hip and trunk strength and its role in controlling multi-directional rotational movement tasks (e.g., pivot) should be conducted to further evaluate training factors that can potentially aid in the reduction of lower extremity injuries. Lastly, the drop-jump is performed in an anticipated fashion. These factors may suggest that the drop-jump task, in a controlled laboratory environment, might not induce sufficient risk/strain to the ACL to allow a clear understanding of the injury mechanism.

Conclusions

Overall, we found that there were differences in kinematic and kinetic variables between the three landing tasks. Particularly, the pivot task exhibited increased knee valgus position and internal varus moment at initial contact and peak stance compared to the sidestep cutting and drop-jump tasks. The pivot also had decreased knee flexion at initial contact and peak stance and increased peak posterior ground reaction force. When combining all the factors, it appears that the athletes presented a more erect posture during the pivot task, and adopted strategies that may place higher loads on the knee joint, and increase the strain on the ACL. Studying the female population in isolation may grant detailed insight into their landing patterns. However, to fully understand the underlying mechanisms causing the gender disparity on injury rates, future studies should include male counterparts, as well as athletes screened and classified at high and low risk for injury. Future studies should focus on the foot-landing strategies' influence on the proximal structures of the lower extremity and differentiate how the movement task (e.g., vertical vs. horizontal) influences the jump-landing movement strategy. The influence of instruction on jump-landing patterns should be further evaluated for various motor tasks to provide evidence-based instructional approaches for injury prevention and investigate how these changes affect performance outcomes. Various approaches (e.g., sagittal vs. frontal plane) to identify the primary risk factor for ACL have been proposed and debated throughout the literature, yet the failure to take into account the task x person x environment trichotomy leads to silo viewpoints that does not account for all the possibilities for ACL injury occurrence. Future assessments should be conducted utilizing a dynamical holistic approach to movement tasks to better account for all factors that can influence movement patterns and thus act as potential risk factors for injury.

CHAPTER V

Experiment III – Biomechanical Differences Between Tasks and Foot Landing Techniques

Title: Biomechanical Differences Between Tasks and Foot Landing Techniques

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Introduction

Injury to the Anterior Cruciate Ligament (ACL) is a common knee injury, often resulting in potential long-term effects for the injured individual (Griffin, et al., 2006; Hootman, Dick, & Agel, 2007; Lohmander, et al., 2004). The injury is more commonly observed during deceleration phase of the action, particularly when coupled with a change of direction (Boden, et al., 2000b; Olsen, et al., 2004). The likelihood for injuries from such motions tends to be more highly associated with female athletes. Indeed the chance of injury for females, has been reported to be at 2 to 9 times greater compared with their male counterparts (Griffin, et al., 2006). Recent ACL Research Retreat consensus meetings have theorized that a non-contact ACL tear is multi-factorial in nature along four risk factor categories: environmental, neuromuscular, anatomical, and hormonal (Griffin, et al., 2000; Griffin, et al., 2006; Shultz, Schmitz, & Nguyen, 2008). Of these four factors, it has been proposed that the two most likely to be amenable to modification are those risk factors which are neuromuscular and/or biomechanical in nature (Caraffa, Cerulli, Projetti, Aisa, & Rizzo, 1996; Chappell & Limpisvasti, 2008; DiStefano, Padua, DiStefano, & Marshall, 2009; Gilchrist, et al., 2008; Giza, Silvers, & Mandelbaum, 2005; Herman, et al., 2009; Hewett, et al., 1999; Hewett, Stroupe, Nance, & Noyes, 1996; Holm, et al., 2004; Onate, et al., 2005; Petersen, et al., 2005). Despite the design of numerous interventions specifically tailored to address neuromuscular and biomechanical factors, the injury ratio has remained steady over the past decade, with a female-male injury ration of 2.38 in 1995 and a 2.75 in 2005 (Agel, et al., 2005; Arendt & Dick, 1995; Mountcastle, Posner, Kragh, & Taylor, 2007). The critical issue that still needs to be addressed is the cause-and-effect relationship between neuromechanical risk factors and the potential for non-contact ACL injury.

Recently, researchers have focused on the role of ankle position at initial ground contact as a potential biomechanical marker for ACL injury (Boden, et al., 2009; Cortes, et al., 2007a). Few studies have investigated the contribution of foot landing technique to the neuromuscular and biomechanical aspects during landing. In one study, Kovacs and colleagues (1999) did not find any difference in knee flexion at initial contact between the techniques while evaluating the effects of two landing techniques (heel and toe landing). Interestingly, the authors reported that during the heel landing the hip and knee flexed (range of motion) 1.2 and 2 times more than during the toe landing. Combined with a 3 times higher ground reaction force in the heel landing, this fact can suggest that the participants had to increase the range of motion to accommodate the increased forces experienced with heel landing.

Cortes and associates (2007) investigated the effects of landing techniques on biomechanical factors and gender. They reported that there was no difference between male and female for the various foot positions; however, they did report a significant difference between landing techniques in some of the theorized risk factors. During the rearfoot landing technique, the subjects had approximately 2.5 times greater landing forces than during the forefoot and self-preferred landing techniques, similar to the findings of Kovacs et al. (1999). Simultaneously, both male and female participants were in a more erect lower extremity position (e.g., lack of knee flexion) while performing the rearfoot landing technique than during the two other techniques. These two factors combined with a minimal time to maximum vertical ground reaction force, makes it plausible to suggest that the participants are placing greater demands on the knee joint when using a heel-to-toe strategy, thus minimizing the force absorption by the ankle joint and calf musculature. Supporting this concept is the observational study conducted by Boden and Hewett (2009). The authors analyzed videos of non-contact ACL injuries during basketball games. They reported that injuries commonly occurred with the ankle in a dorsi-flexion (heel contact) strategy at initial ground contact.

Recently, the two landing strategies (e.g., forefoot and rearfoot) have been quantified during a drop-jump task (Cortes, et al., 2007a; Kovacs, et al., 1999). This task provides baseline information on the different biomechanical parameters and its constraints can be easily controlled. Athletic tasks, however, (i.e., pivot and sidestep cutting) presented in an unanticipated fashion that require increased attention demands and reaction to a stimulus may provide further understanding on how the biomechanical parameters interrelate with various foot landing techniques.

A commonly observed motion during a soccer game is a cutting motion from an opponent. It is reported that this motion increases the likelihood of injury (Boden, et al., 2000b). Researchers have attempted to replicate this motion within a laboratory environment using a sidestep cutting task (McLean, Huang, et al., 2004; Pollard, et al., 2007), and a pivot task with a 180 degrees change in direction (Greig, 2009; McLean, Walker, et al., 2005). There are some natural differences between these tasks. The 180-degree maneuver oftentimes seen in soccer requires a complete deceleration with a change in direction followed by acceleration to maximum speed and was reported to provide a more realistic representation of a soccer task (Greig, 2009). These inherent differences suggest that the control mechanism and demands between these tasks are different. Thus, the multiple biomechanical risk factors may play a different role depending on the task, as it may require different demands from the motor system (Newell & Slifkin, 1998b). Few studies have attempted to quantify and compare biomechanical parameters among tasks and landing techniques.

The purpose of this study is to evaluate the effects of two landing techniques (rearfoot and forefoot) in biomechanical risk factors (knee and hip flexion, knee valgus, hip and knee rotation, knee flexion and valgus moment, and ground reaction forces) while performing two unanticipated tasks (sidestep cutting and pivot). We hypothesize that the rearfoot landing technique and the pivot task will produce significantly distinct kinematic and kinetic parameters than the forefoot landing technique and sidestep cutting task.

Methodology

Participants. Twenty female collegiate soccer athletes (age = 20 ± 0.9 years old; height = 1.67 ± 0.05 cm; mass = 63.2 ± 10.1 kg) from a Division I institution were chosen to participate in this study. Participants' numbers were based upon *a priori* power calculations (Greig, 2009; McLean, Huang, et al., 2005). Participants were screened to ensure none had any previous hip, low back, knee, or severe ankle injuries within the last six months or surgeries within the last 2 years. The dominant leg, defined as the leg that the participant would use to kick a soccer ball as far as possible, was used for analysis. Prior to data collection, approval of the research through Institutional Review Board, and written informed consent for all participants was obtained.

Experimental Procedure. Participants wore spandex shorts, sports bra and the team running shoes (Adidas Supernova, AG, Herzogenaurach, Germany). Participants completed a 5-minute cycling warm-up and 5-minute self-directed stretching, prior to data collection. General anthropometric measures were taken for each participant. The same researcher completed all anthropometric measures. After the warm-up, reflective markers were placed on specific body landmarks according to a modified Helen Hayes marker set (Kadaba, et al., 1990). Participants were required to partake in two different landing techniques (forefoot and rearfoot) while performing two tasks (sidestep and pivot).

A Brower timing system (Brower Timing Systems, Draper UT, USA) was used to control the approach speed. Participants had to attain a minimal speed of 3.5 $m.s^{-1}$. The participants stood on the beginning of a running platform pressing a pad to trigger the speed-timing device. For the sidestep cutting task, the participants started running, and stepped with the dominant foot on the force plate. At that moment they had to perform a cutting motion to the contra-lateral side of the dominant foot touching the force plate. The running pathway was constrained to 35 to 55 degree angle to provide an optimal cutting angle of 45-degrees (Colby, et al., 2000). For the pivot task, the participant ran and planted onto the force plate with the dominant foot and pivoted 180 degrees and ran to the starting position (Greig, 2009). Participants were permitted three practice trials for each technique. The forefoot landing consisted of initial contact with the toes first on the force plates followed by the rearfoot. For the rearfoot landing, the initial contact was with the heels first on the force plates followed by the forefoot. The forefoot and rearfoot landing techniques were counterbalanced between subjects. Five successful trials were presented in an unanticipated fashion and collected for each task. If participants did not place the dominant foot on the force plate, lost balance, or did not perform the appropriate task based on the cue generated by the software the trial was not successful and discarded from analysis. There was a one-minute rest period between trials to minimize fatigue.

Instrumentation. Kinematic measures of the lower extremity were captured using an eight-camera high-speed (500-Hz) motion capture system (Vicon Motion Systems Ltd., Oxford, England). Reflective markers placed on specific body landmarks (anterior and posterior iliac spine, thigh, knee, shank, malleoli, heel, and fifth metatarsal) were tracked via the motion capture system. A standing (static) trial with the participants standing on the force plates with shoulders abducted at 90 degrees was obtained. From the standing trial a kinematic model (pelvis, thigh, shank, and foot) was created for each participant using Visual 3D software (C-Motion, Inc, Germantown MD, USA), using a least-squares optimization. (Lu & O'Connor, 1999) This kinematic model was used to quantify the motion at the hip, knee, and ankle joints. A Cardan angle sequence was used to calculate joint kinematics. Marker trajectories were filtered with a 4^{th} order low-pass Buttherworth filter with a cutoff frequency of 7Hz, whereas ground reaction force data were filtered with a fourth-order low-pass Buttherworth filter with a cutoff frequency of 25Hz. A standard inverse dynamics analysis was employed to the kinematic and ground force data to calculated 3-D forces and moments(Winter, 2005). Segment inertial characteristics were estimated for each participant as per the methods of Dempster (Dempster, 1955). Intersegmental joint moments are defined as internal moments (e.g., a knee internal extension moment will resist a flexion load applied to the knee).

Data Analysis. All data were reduced using Matlab 6.1 (The MathWorks, Inc, Natick MA, USA) software with the creation of a custom made program to export into a Microsoft Excel spreadsheet the variables of interest. Each of the five trials were averaged and exported into SPSS version 16.0 (SPSS Inc, Chicago IL, USA) for data analysis. The alpha level was set *a priori* at 0.05. Separate repeated measures analyses of variance (ANOVA) with landing technique (2) and task (2) as the repeating factors, were conducted to evaluate the kinematic (hip flexion, knee flexion, and knee valgus) and kinetic (vertical and posterior ground reaction forces, knee flexion-extension and abduction-adduction moments, and ankle plantarflexion-dorsiflexion moments) parameters at different time instants (initial contact, peak vertical ground reaction force, and peak posterior ground reaction force).

Results

Descriptive statistics with means, standard deviations, and 95% confidence intervals are presented in Table 1 and Table 2.

Landing Technique

At initial contact, the forefoot technique had significantly higher posterior ground reaction force than the rearfoot technique ($F_{1,18}$=59.217, p<0.001). The forefoot landing technique had significantly higher knee flexion than the rearfoot ($F_{1,18}$=28.294, p<0.001), knee flexion moment ($F_{1,18}$=11.853, p=0.003), and knee adduction moment ($F_{1,18}$=32.645, p<0.001) at initial contact. The rearfoot landing technique had significantly higher hip flexion than the forefoot ($F_{1,18}$=16.002, p=0.001).

At peak vertical ground reaction force, the forefoot landing technique had significantly higher knee flexion than the rearfoot ($F_{1,18}$=18.295, p<0.001). The rearfoot landing technique had significantly higher knee abduction than the forefoot ($F_{1,18}$=5.446, p=0.031). At peak posterior ground reaction force, the forefoot landing technique had increased knee flexion angle when compared with the rearfoot landing technique ($F_{1,18}$=23.540, p<0.001). At peak stance, the forefoot landing technique had significantly higher knee abduction than the rearfoot ($F_{1,18}$=6.735, p=0.018).

Task

At initial contact, the sidestep task had significantly higher knee flexion at initial contact than the pivot ($F_{1,18}$=149.999, p<0.001), and higher hip flexion ($F_{1,18}$=4.984, p=0.039). During the pivot task, participants had increased knee abduction angle when compared with the sidestep cutting ($F_{1,18}$=36.361, p<0.001), and higher knee adduction moment ($F_{1,18}$=5.971, p<0.001).

At peak vertical ground reaction force, the sidestep cutting task had significantly higher knee flexion than the pivot ($F_{1,18}$=46.964, p<0.001), and the pivot task had higher knee abduction than the sidestep cutting task ($F_{1,18}$=54.713, p<0.001). At peak posterior ground reaction force, the pivot task created significantly higher posterior ground reaction force than the sidestep cutting ($F_{1,18}$= 53.900, p<0.001). The sidestep cutting task had increased knee flexion angle than the pivot task ($F_{1,18}$=33.472, p<0.001). At peak stance, the pivot task had significantly higher hip flexion than the sidestep cutting task ($F_{1,18}$=11.260, p=0.004), as well as higher knee abduction ($F_{1,18}$=35.932, p<0.001), and higher knee adduction moment ($F_{1,18}$=9.304, p=0.007). Lastly, the sidestep cutting had significantly higher knee extension moment than the pivot task ($F_{1,18}$=103.668, p<0.001).

Landing Technique x Task

At initial contact, the forefoot landing technique had significantly higher posterior ground reaction force during the sidestep when compared with the rearfoot landing technique, and when compared with the rearfoot landing technique during the pivot task ($F_{1,18}$=36.854, p<0.001). The forefoot landing technique during the pivot task had increased knee abduction than during the sidestep with the forefoot, as well as compared with the sidestep and pivot when using the rearfoot landing technique ($F_{1,18}$=39.636, p<0.001). The pivot task during the forefoot landing technique had increased knee flexion moment when compared to the sidestep and pivot tasks during the rearfoot landing technique ($F_{1,18}$=5.573, p=0.03). The pivot task during the forefoot landing technique had increased knee adduction moment when compared with the sidestep task using the forefoot landing technique, and the pivot and the sidestep using the rearfoot ($F_{1,18}$=11.200, p=0.004).

At peak vertical ground reaction force, the rearfoot landing technique during the sidestep task had increased knee flexion than the pivot with the both landing techniques ($F_{1,18}$=14.311, p=0.001). The pivot task using the forefoot landing technique had increased knee abduction angle than using the rearfoot and than the sidestep with both landing techniques; the pivot with the rearfoot also had significantly higher knee abduction than the sidestep with rearfoot and forefoot landing techniques ($F_{1,18}$=33.458, p<0.001).

Discussion

The present study was designed to evaluate kinematic and kinetic differences between two commonly used landing tasks – sidestep cutting and pivot – while using two different landing techniques – forefoot and rearfoot. One of the main results to emerge from this study is that the pivot task has distinct neuromechanical characteristics than sidestep cutting task. This result was observed when the pivot task was performed with either one of the landing techniques. Specifically, the rearfoot landing technique had increased knee valgus angles, decreased knee flexion angle, and knee flexion moment when performed with the pivot task. Regardless of the landing technique, the pivot task also presented increased knee valgus angles and moments, and a decrease in peak posterior ground reaction force with a decreased knee flexion angle. The demarcation of the tasks with the different landing techniques may possibly suggest that the combination of task and landing technique present differentiated characteristics and that the injury mechanism may be dependent on the combination of these two factors.

We found that the pivot task had increased posterior ground reaction, combined with decreased knee flexion. An increased posterior ground reaction force has been theorized to increase the strain on the knee ligamentous structures by enhancing the proximal anterior tibia shear force (Sell, et al., 2007; Yu, et al., 2006). Proximal anterior tibia shear force is believed to create an anterior displacement of the tibia, thus increasing the strain on the ACL (Sell, et al., 2007; Yu, et al., 2006). The low knee flexion angles presented at peak ground reaction force and the increased force during the pivot task might be augmenting the quadriceps activation (Blackburn & Padua, 2009). It is worth noting that this became especially relevant when the pivot task was performed with the rearfoot landing technique. When the pivot task was completed with rearfoot landing technique, the participants barely achieved 30 degrees of knee flexion, whereas with the forefoot technique the obtained knee flexion angle was well above 30 degrees for both tasks.

Low knee flexion angles during the first 50% of the stance phase have been theorized to raise the likelihood of injury (Wojtys, et al., 2002), since the strain placed by the quadriceps on the ACL can potentially cause tears (Nisell, 1985). The maximum knee flexion across tasks and techniques ended in similar knee flexion angles, however, at the time instants theorized to increase ACL strain (e.g., initial contact, peak posterior and vertical ground reaction forces) our participants had diminished knee flexion angle, which may not be sufficiently protective of the knee structures. These results are similar to those reported by Cortes and colleagues (2007). The authors found that knee flexion angle was significantly lower at peak vertical ground reaction force for the rearfoot landing technique when compared with the forefoot technique during a drop-jump. We found a decreased knee flexion angle for the rearfoot while performing the pivot task, but not during the sidestep task. It appears that the neuromechanical demands of a pivot task are placing the participants at increased risk of straining their ACL, especially when performing this task with a rearfoot landing strategy.

Surprisingly, we did not find a difference in vertical ground reaction between landing techniques. Previous research (Kovacs et al., 1999), previously reported that the vertical ground reaction force was up to 3.4 times greater during the rearfoot than the forefoot landing technique. However, the discrepancy between the two studies is probably a result of different methodological practices. Kovacs utilized a vertical drop from a box, whereas our task required mainly horizontal momentum with minimal vertical motion. In our study, the participants had to make contact with the force plates and perform i) a deceleration with rapid acceleration and change of direction (sidestep), and ii) a complete deceleration combined with 180 degrees change of direction followed by an acceleration phase (pivot). The difference between task demands, regardless of the landing technique used, may possibly explain the lack of difference in vertical ground reaction force. This is further supported with the posterior ground reaction force (horizontal force) results, where the primary difference between tasks and landing techniques occurred. Several studies and concepts of motor control support the notion that the multiple risk factors can vary with different task constraints (Landry, et al., 2007; Newell, 1996; O'Connor & Bottum, 2009).

The first key aspect is the finding that the participants were always in a valgus position irrespective of the task and/or technique used. However, they were at increased valgus while performing the pivot task. During the sidestep task the valgus angles were smaller, and in some time instants, almost close to neutral position (i.e., initial contact during the sidestep combined with the forefoot landing technique). Previous studies, have reported through visual observation that participants were in valgus position at the time of injury (Boden, et al., 2000b; Krosshaug, et al., 2007b). Recently, Boden and colleagues (2009) reported that during ACL injury events, the athletes were in a heel contact position (rearfoot) at time of ground contact. The valgus position presented by our participants could be theorized as an increased risk for injury, however, when carefully looking at the values this can be rather puzzling. During the sidestep cutting, the knee valgus angle increased with use of the rearfoot landing technique, whereas the knee valgus decreased with the rearfoot landing technique during the pivot task. A possible explanation for this fact is that the participants were in a more extended knee position while performing the pivot task with the rearfoot landing technique during the 180-degree change in direction. Although they kept similar knee flexion angles during the sidestep task, with the demands of the rearfoot landing technique, their dominant knee may exhibit increased valgus displacement during the rapid deceleration-acceleration phase with a 45-degree change of direction. These factors (pivot task and rearfoot landing technique), combined with the increased internal adduction moment during the pivot task may potentially increase the stress on the knee structures, particularly the ACL. Several authors have theorized that an increase in adduction loading seems to be a strong predictor for increased risk of ACL injuries (Hewett, et al., 2004; McLean, Huang, et al., 2004; McLean, Huang, et al., 2005). Thus, developing individualized intervention strategies that focus on minimizing knee valgus position and loading, combined with proper landing technique (i.e., forefoot landing technique) have been proposed (Hewett, Myer, & Ford, 2005; Hewett, Myer, Ford, et al., 2005; Renstrom, et al., 2008).

Lastly, our participants presented increased hip flexion during the pivot task with rearfoot landing technique. Previous studies focusing on the effects of landing techniques found similar results (Cortes, et al., 2007a; Kovacs, et al., 1999). The authors reported that hip flexion angles increased when using a rearfoot landing technique, possibly to compensate for the decreased action of the plantar-flexors and knee flexion angles. Previous studies have also observed increased hip flexion in athletes that tore their ACL during a game (Boden et al., 2009; Krossaugh et al., 2007). A rationale is still to be proposed about ACL tear events occurring with increased hip flexion. Further research related to the neuromuscular contribution of the hip musculature seems necessary, especially when performing tasks and landing techniques that are theorized to increase the stress on the knee joint (i.e., pivot task with rearfoot landing technique).

Conclusions

Overall, the results of this study highlighted that there were inherent differences in biomechanical outcomes between foot-landing techniques and tasks, as well as an interaction of both. Specifically, the pivot task had increased knee valgus angles and internal varus moments, decreased knee flexion angles, and increased posterior ground reaction forces when using the rearfoot landing technique, as well as increases hip flexion angles with rearfoot landing technique. The decreased knee flexion angle associated with higher posterior ground reaction force can potentially be creating higher stress and strain on the ACL. This force has been positively correlated with proximal anterior tibia shear force and tibia strain. The small knee flexion angle may not allow for proper hamstring activation to protect the tibia from anterior displacement and decrease ACL strain. In the course of prevention programs, attention should be taken to provide feedback regarding the use of proper landing technique, given that the rearfoot has been observed during ACL tear events (Boden, et al., 2009). The consideration of task and person interaction is also fairly important in the multifactorial approach to ACL injury prevention since we have shown that different inherent task demands can reflect significant movement differences. Additionally, future research considering the ecological validity of the environmental interaction on person and task movement pattern outcomes must be considered as well when constructing injury prevention assessment and training paradigms.

CHAPTER VI

CONCLUSIONS

Overall, the three experiments have provided some more insight into landing characteristics, and how tasks and landing techniques provide differentiated mechanics. In the first experiment, we found that the kinematic variables were shown to have a stronger coupling relationship with both tasks. In addition, the drop jump was characterized by different variables when compared to the sidestep cutting. This cutting task displayed increased variability in hip, knee, and ankle flexion. The second experiment determined differences in kinematic and kinetic variables between three landing tasks (drop-jump, pivot, and sidestep cutting). Particularly, the pivot task exhibited increased knee valgus position and internal varus moment at initial contact and peak stance compared to the sidestep cutting and drop-jump tasks. The pivot also had decreased knee flexion at initial contact and peak stance and increased peak posterior ground reaction force. In our last experiment, we quantified the biomechanical effects of foot-landing techniques during two unanticipated athletic tasks. We found there were inherent differences in biomechanical outcomes between foot-landing techniques and tasks, as well as an interaction of both. Specifically, the pivot task had increased knee valgus angles and internal varus moments, decreased knee flexion angles, and increased posterior ground reaction forces when using the rearfoot landing technique, as well as increased hip flexion angles.

These experiments have illustrated the influence of biomechanical parameters in specific tasks; how those tasks differ from each other; and the interaction of athletic tasks with different foot landing techniques. Based on all the three experiments, there is indication that the mechanism of injury is potentially different between a vertically oriented drop jump as compared to a sidestep cutting movement that requires rapid horizontal oriented deceleration/acceleration combined with change of direction. The increased variability in the sidestep task can be seen as a protective mechanism due to the adaptability of the participants to the increased challenge demands of the sidestep task. The athletes presented a more erect posture during the pivot task, and adopted strategies that may place higher loads on the knee joint, and increase the strain on the ACL. This was further observed when performing the pivot task with the rearfoot landing technique.

The influence of instruction on jump-landing patterns should be further evaluated for various motor tasks to provide evidence-based instructional approaches for injury prevention, and to investigate how these changes affect performance outcomes. Various approaches (e.g., sagittal vs. frontal plane) to identify the primary risk factor for ACL have been proposed and debated throughout the literature, yet the failure to take into account the task x person x environment trichotomy leads to silo viewpoints that does not account for all the possibilities for ACL injury occurrence. In the course of prevention programs, attention should be taken to provide feedback regarding the use of proper landing technique, given that the rearfoot approach has been observed during ACL tear events (Boden, et al., 2009). Lastly, future studies analyzing the interaction of task and landing technique under fatigue conditions with decision-making processes involved (i.e., unanticipated) needs to be explored.

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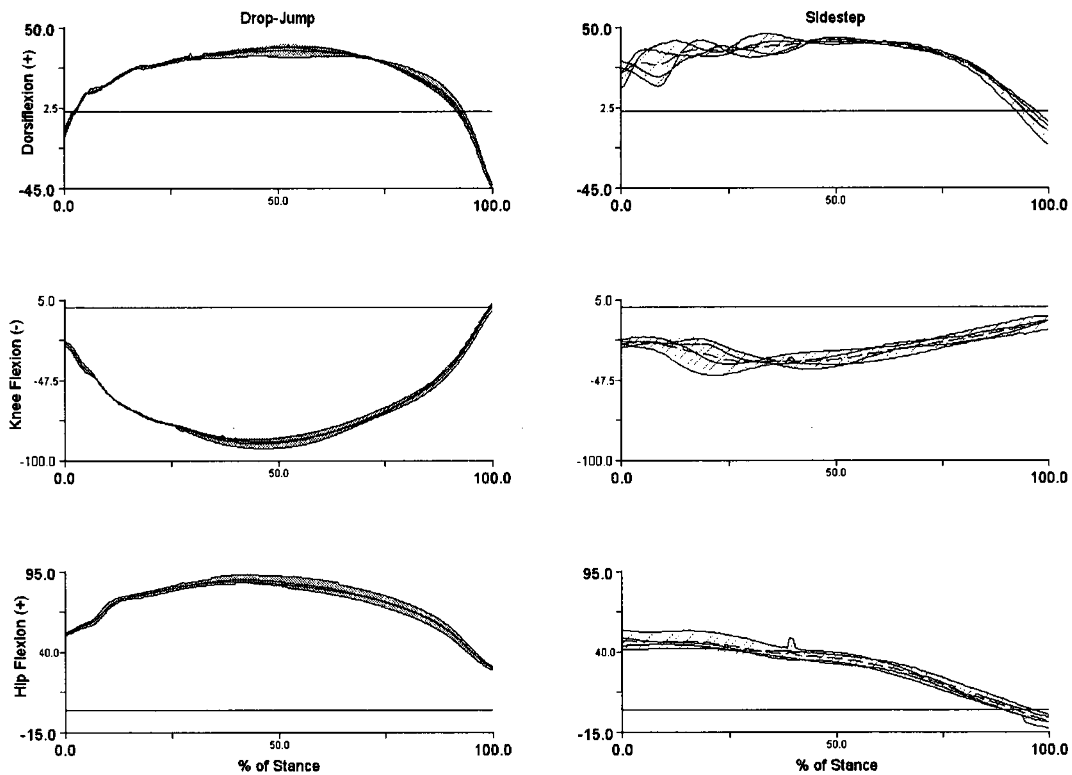
APPENDIX I - Figure 3.1 Typical curves of flexion angles (ankle, knee, and hip) for the drop jump and sidestep cutting tasks



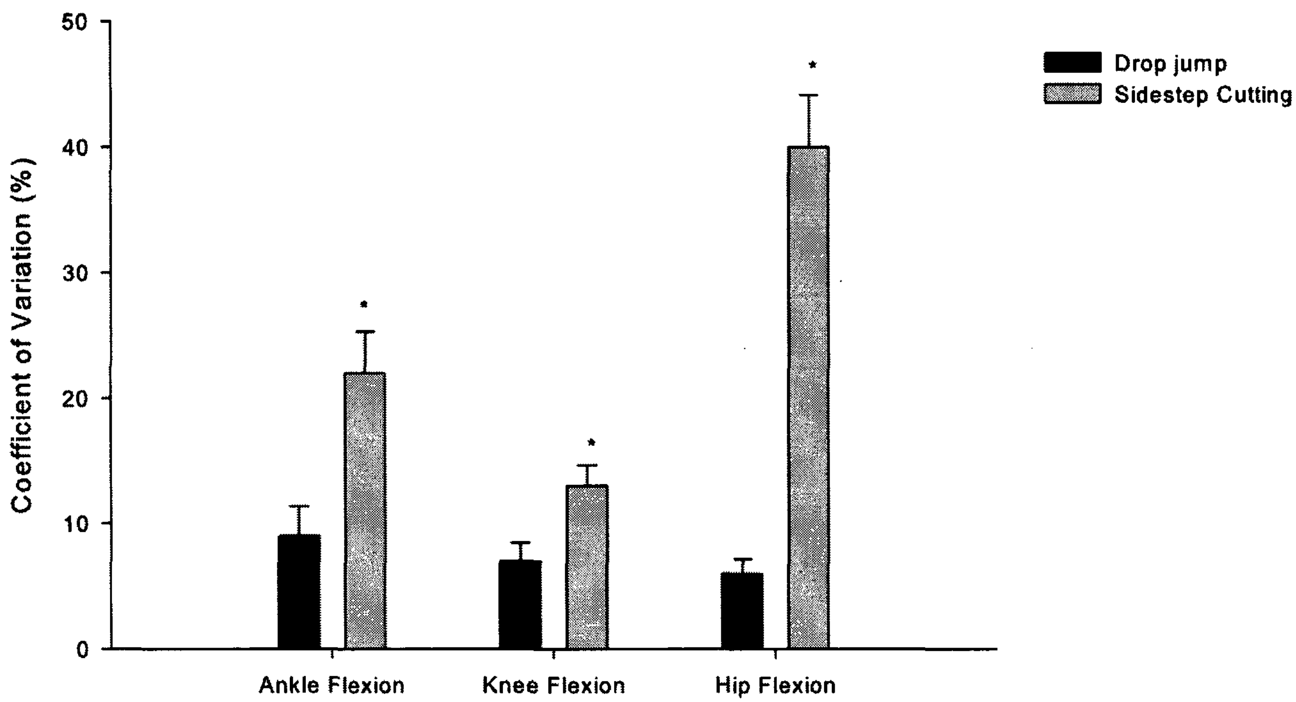
Appendix II - Table 3.1. Eigenvalues (EV), % of variance, and correlation coefficients between the variables for principal component 1 for the drop-jump and sidestep cutting

Task	Variables	Correlation Coefficient	Eigenvalues	Percent of variance Explained (%)
			15.340	23.24%
Drop-Jump				
	Trunk Flexion at Peak Posterior Ground Reaction Force	.974		
	Trunk Flexion at Peak Knee Shear Force	.971		
	Trunk Flexion at Peak Vertical Ground Reaction Force	.970		
	Trunk Flexion at Initial Contact	.970		
	Trunk Flexion at Peak Knee Flexion	-.818		
	Knee Valgus at Peak Vertical Ground Reaction Force	.819		
	Knee Valgus at Initial Contact	.617		
Sidestep cutting			11.851	18.23%
	Hip Flexion Moment at Peak Posterior GRF	-.757		
	Knee Flexion Moment at Peak Knee Shear Force	.872		
	Knee Shear Force at Peak Knee Shear Force	.828		
	Ankle Flexion at Initial Contact	.655		
	Ankle Flexion at Peak Knee Flexion	.613		
	Ankle Flexion at Peak Knee Shear Force	.878		

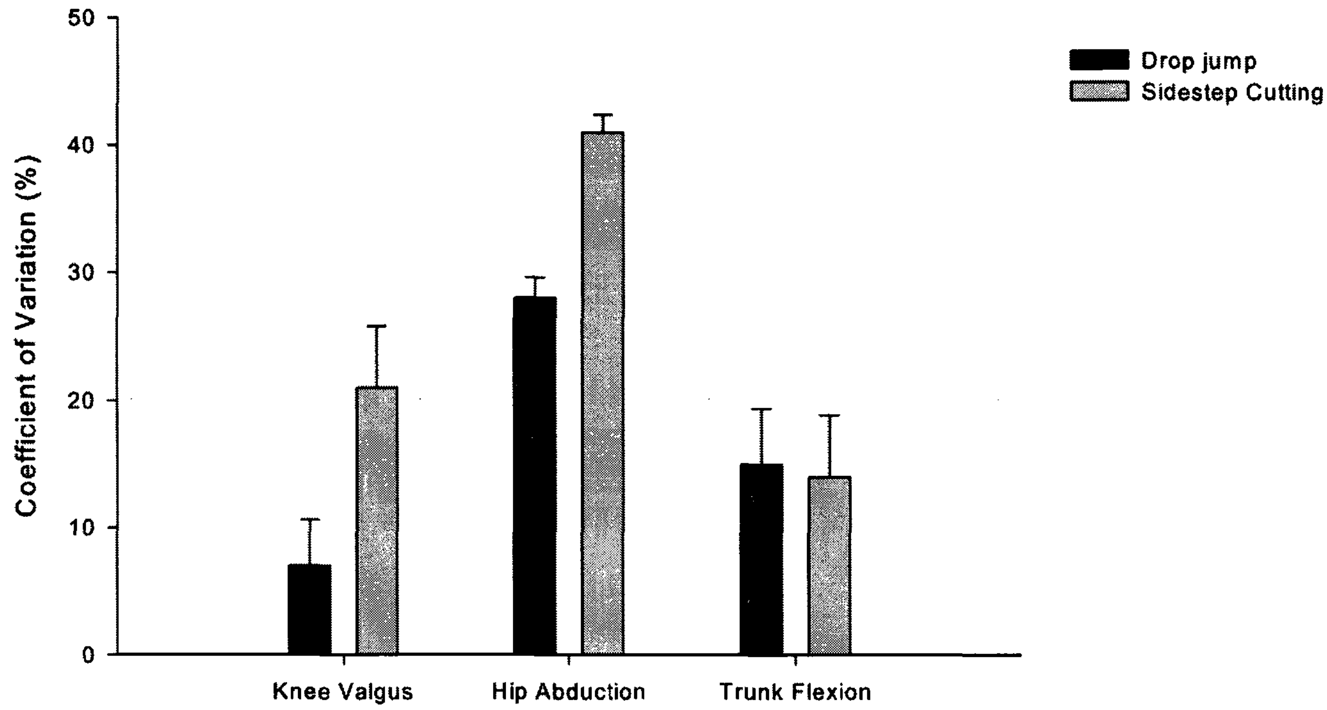
APPENDIX III - Table 3.2. Eigenvalues (EV), % of variance, and correlation coefficients between the variables for principal component 2 for the drop-jump and sidestep cutting

Task	Variables	Correlation Coefficient	Eigenvalues	Percent of variance Explained (%)
Drop-Jump			10.090	15.29%
	Hip Abduction at Initial Contact	-.832		
	Hip Abduction at Peak Posterior GRF	-.785		
	Hip Rotation at Peak Posterior GRF	.643		
	Knee Valgus at Peak Posterior GRF	.633		
	Knee Valgus at Initial Contact	.632		
	Knee Valgus at Moment Peak Posterior GRF	-.612		
	Ankle Flexion at Peak Knee Shear Force	.641		
	Ankle Flexion at Peak Posterior GRF	.860		
	Ankle Flexion at Peak Knee Flexion	.749		
Sidestep Cutting			10.254	15.78
	Trunk Flexion at Initial Contact	.779		
	Maximum Hip Flexion	.617		
	Knee Flexion at Peak Posterior GRF	.831		
	Knee Flexion at Peak Vertical GRF	.751		
	Ankle Flexion at Peak Vertical GRF	-.892		
	Ankle Flexion at Peak Posterior GRF	-.845		

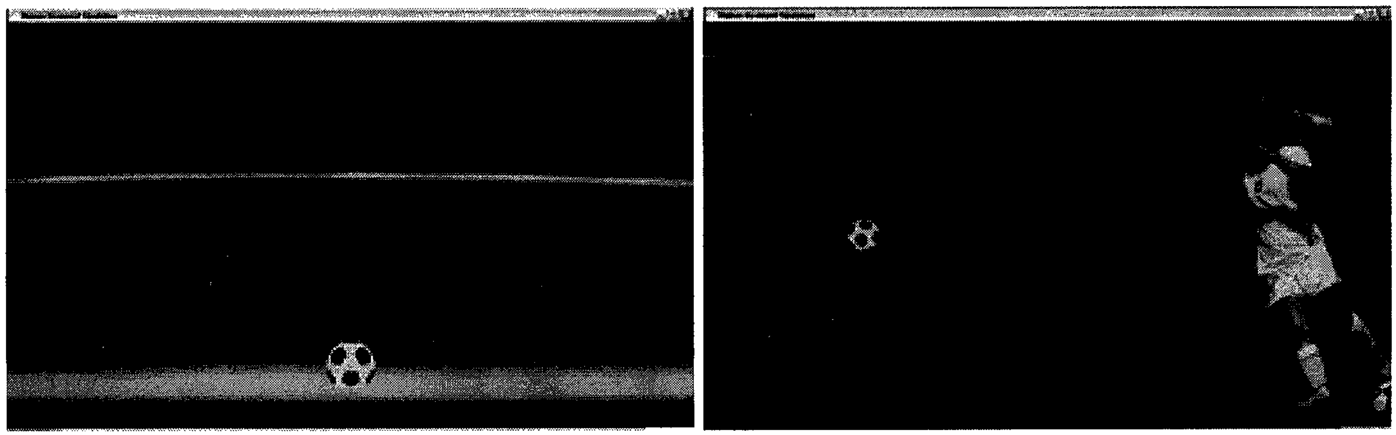
Appendix IV - Figure 3.2 Coefficient of variation between a drop-jump and sidestep cutting for hip, knee, and ankle flexion. Significant level at P<0.05 (*)



Appendix V - Figure 3.3 Coefficient of variation between a drop-jump and sidestep cutting for knee valgus, hip abduction, and trunk flexion



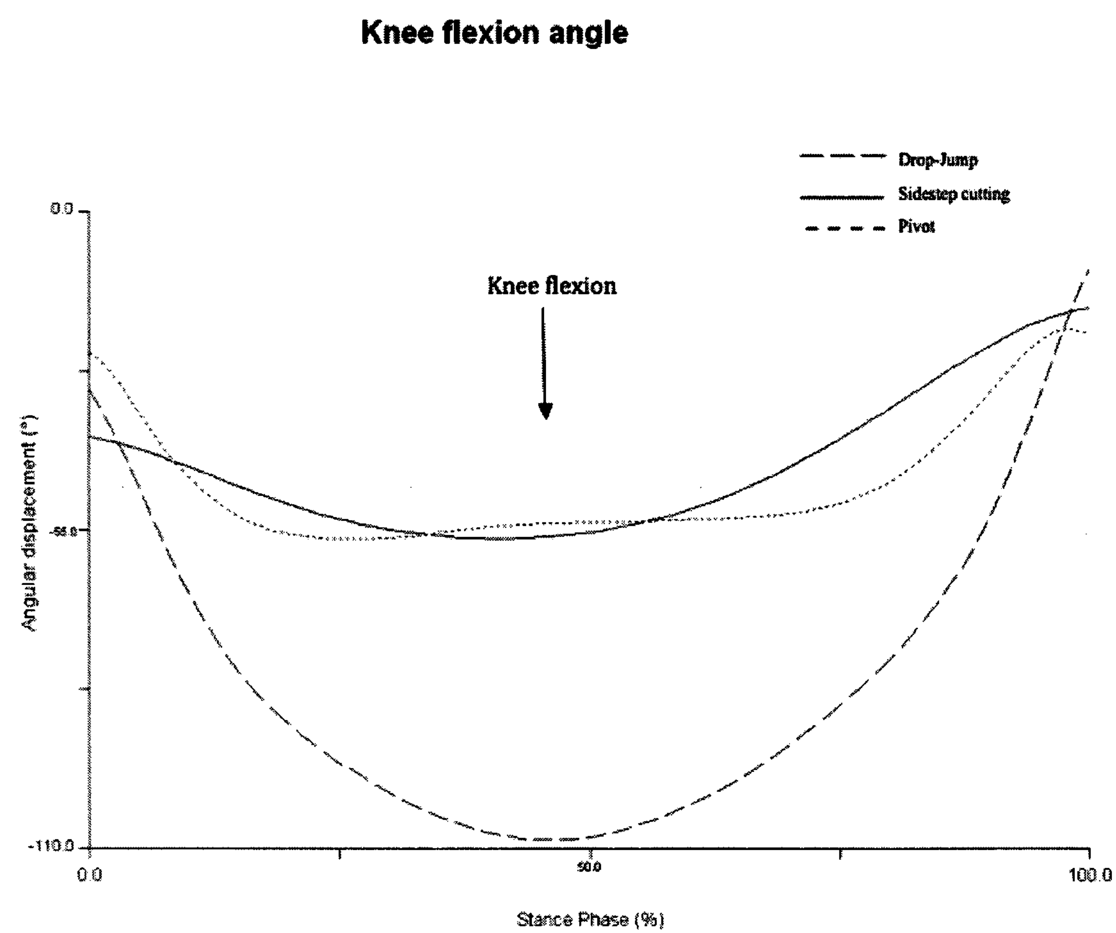
Appendix VI - Figure 4.1 Unanticipated scenario projected onto a screen to mimic a soccer game situation



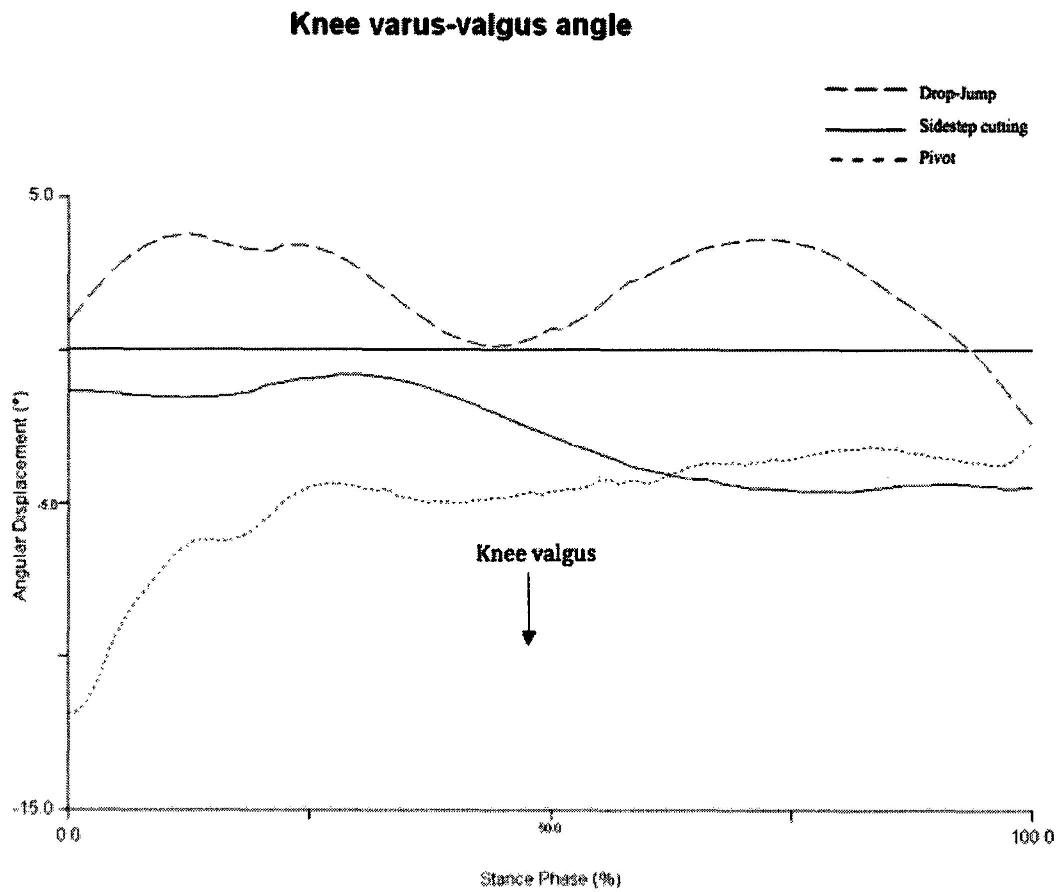
Appendix VII - Table 4.1. Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinematic variables at initial contact, peak stance, and peak vertical ground reaction force (PVGRF) during three athletic tasks

	Drop-Jump				Sidestep cutting				Pivot			
	Mean	SD	95%CI-LB	95%CI-UB	Mean	SD	95%CI-LB	95%CI-UB	Mean	SD	95%CI-LB	95%CI-UB
Initial contact												
Knee flexion	-30.3	5.2	-32.8	-27.8	-38.8	8.4	-42.9	-34.8	-24.3	5.7	-27.1	-21.5
Knee valgus	0.8	4.7	-1.4	3.1	-1.4	9.3	-5.9	3.1	-11.6	6.7	-14.9	-8.4
Hip flexion	53.5	8.3	49.5	57.5	48.6	13.5	42.1	55.1	48.3	8.4	44.3	52.4
Ankle flexion	-6.7	7.4	-10.1	-2.9	-0.8	11.7	-6.4	4.8	2.5	36.4	-15.1	20.0
Peak Stance												
Knee flexion	-108.8	13.1	-115.2	-102.5	-56.5	7.2	-59.9	-53.0	-57.6	7.9	-61.4	-53.8
Knee valgus	-3.9	8.0	-7.8	-0.8	-3.8	10.0	-8.6	1.0	-12.2	7.0	-15.5	-8.8
Hip flexion	95.0	14.7	87.9	102.1	49.1	13.7	42.5	55.7	64.9	13.4	71.3	58.4
PVGRF												
Knee flexion	-73.2	2.7	-85.6	-60.8	-53.9	9.4	-58.4	-49.3	-41.2	8.8	-45.7	-37.0
Knee valgus	3.7	6.4	0.6	6.7	-2.9	10.0	-7.8	1.8	-7.6	10.1	-12.5	-2.8
Hip flexion	75.7	15.6	68.1	83.2	37.0	13.1	30.7	43.3	52.7	11.6	47.1	58.4

Appendix VIII - Figure 4.2 Knee flexion angle during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks



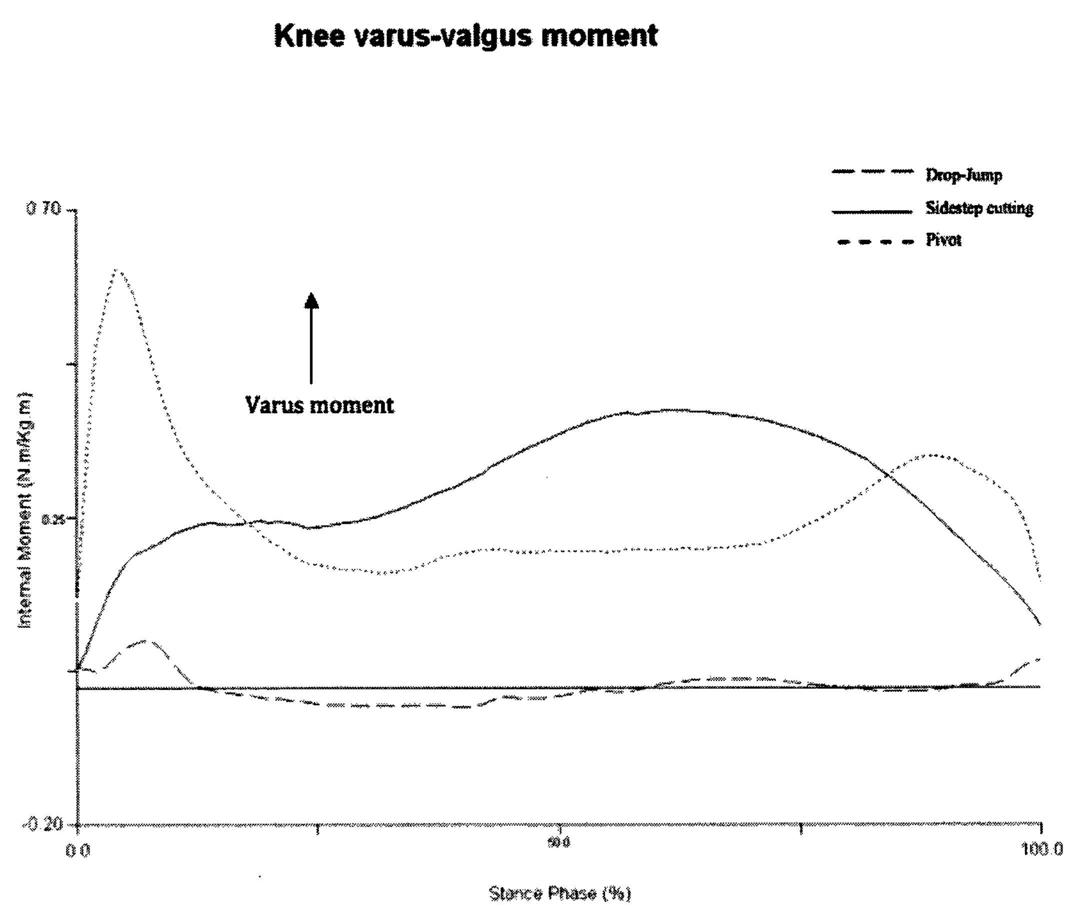
Appendix IX - Figure 4.3 Knee valgus angle during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks



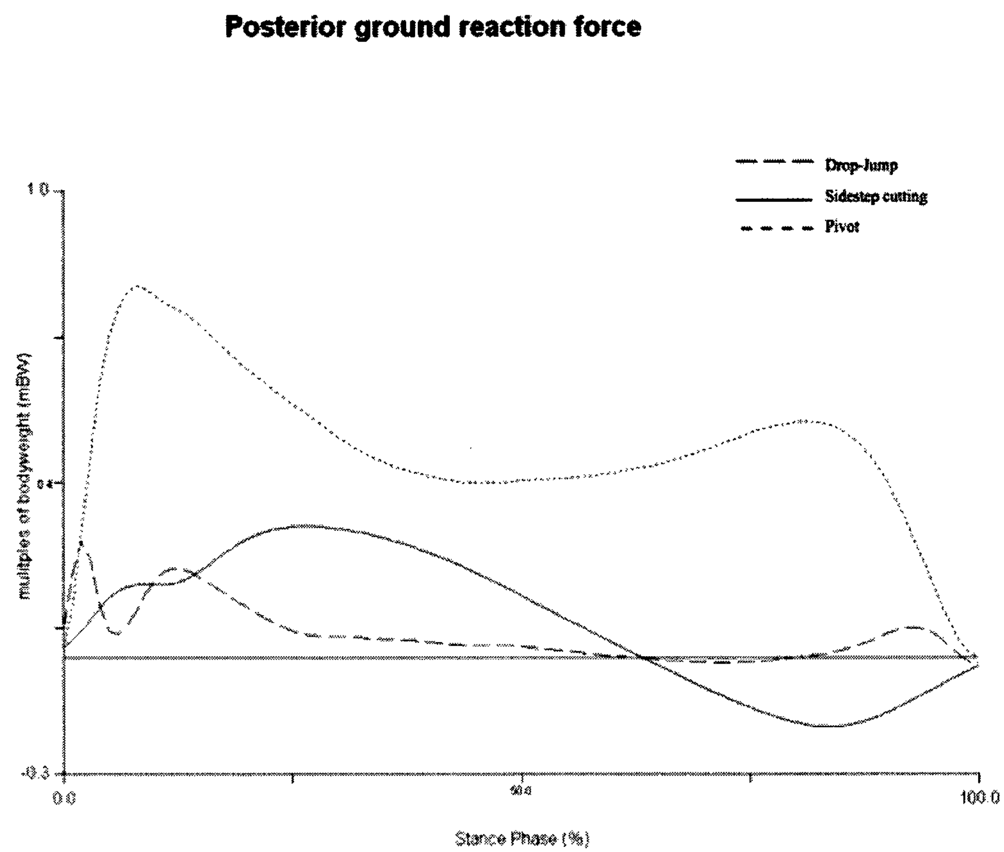
Appendix X - Table 4.2. Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinetic variables at initial contact and peak stance during three athletic tasks

	Drop-Jump				Sidestep cutting				Pivot			
	Mean	SD	95%CI-LB	95%-UB	Mean	SD	95%CI-LB	95%-UB	Mean	SD	95%CI-LB	95%-UB
Initial Contact												
Extension moment	-0.055	0.048	-0.078	-0.031	-0.225	0.066	-0.257	-0.194	-0.266	0.093	-0.311	-0.222
Varus-valgus moment	0.029	0.027	0.016	0.042	0.261	0.044	0.005	0.047	0.128	0.075	0.091	0.164
PGRF	0.068	0.013	0.061	0.074	0.224	0.034	0.006	0.039	0.024	0.386	0.005	0.042
Peak Stance												
Extension moment	0.789	0.208	0.689	0.890	1.131	0.289	1.271	1.150	0.523	0.245	0.405	0.641
Varus-valgus moment	0.140	0.069	0.107	0.173	0.487	0.326	0.330	0.644	0.719	0.300	0.575	0.864
VGRF	1.16	0.16	1.09	1.24	1.64	0.41	1.44	1.84	1.51	0.47	1.28	1.73
PGRF	0.264	0.063	0.233	0.294	0.280	0.121	0.221	0.338	0.801	0.289	0.662	0.941

Appendix XI – Figure 4.4 Knee varus-valgus moment during the stance phase of three tasks – drop-jump, sidestep, and pivot tasks



Appendix XII - Figure 4.5 Posterior Ground Reaction Force during the three landing tasks measured in multiples of bodyweight while performing three tasks – drop-jump, sidestep, and pivot



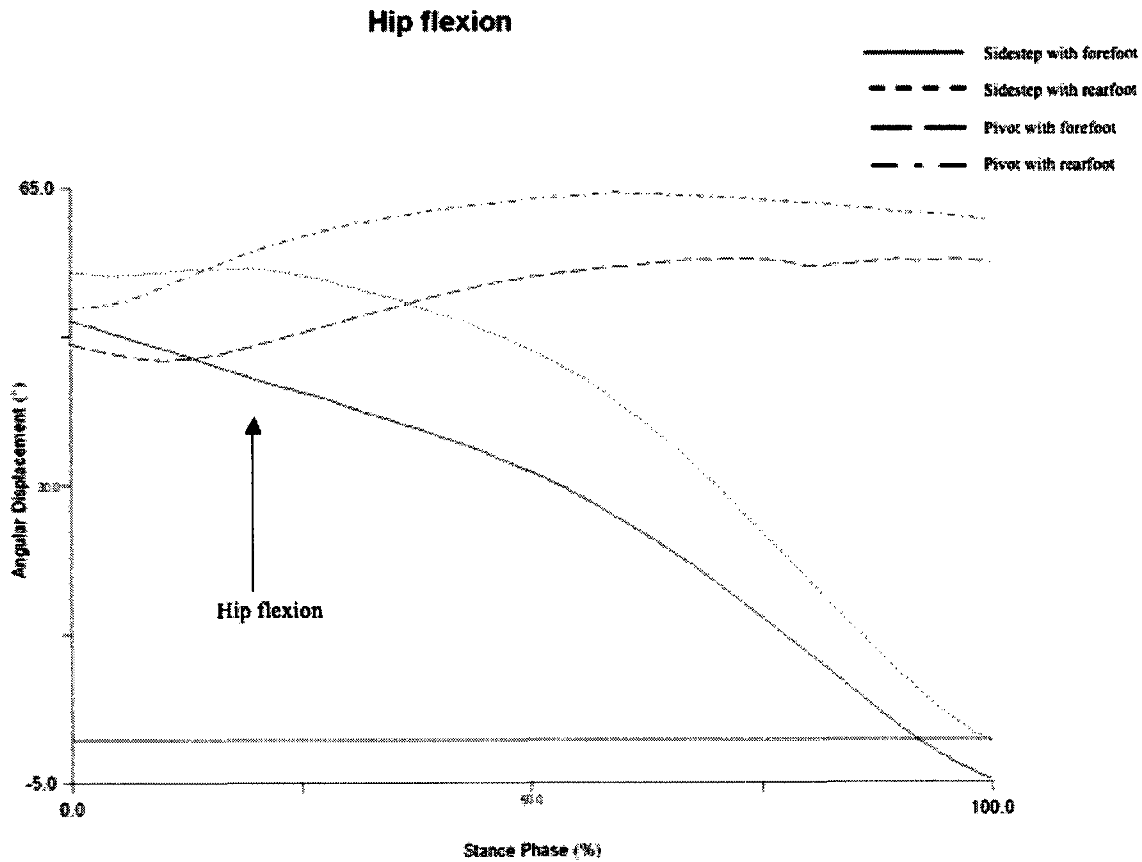
Appendix XIII - Table 5.1. Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinematic variables at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PGRF), and peak stance during two athletic tasks using two landing-techniques

	Sidestep				Pivot			
	Forefoot		Rearfoot		Forefoot		Rearfoot	
	Mean ± SD	CI	Mean ± SD	CI	Mean ± SD	CI	Mean ± SD	CI
Initial contact								
Knee flexion (-) / extension (+)	-42.0±10.3	-37.1, -47	-33.1±6	-30.2, -36.0	-25.9±7.8	-22.1, -29.6	-17.8±5	-15.3, 20.2
Knee abduction (-) / adduction (+)	-0.7±11.2	-6.1, 4.7	-3.1±9	-7.5, 1.2	-12.8±7.9	-16.6, -9.0	-7.4±6.7	-10.7, -4.2
Hip flexion	49.5±10.6	44.4, 56.6	55.1±8.3	51.0, 59.1	46.5±8	42.7, 50.4	50.7±6.9	-47.4, 54.0
PVGRF								
Knee flexion (-) / extension (+)	-48.9±14	-42.1, -55.6	-52.3±7.9	-48.5, -56.2	-40.7±12.5	-34.7, -46.8	-25.6±6.9	-22.3, -28.9
Knee abduction (-) / adduction (+)	-1.4±10.7	-6.5, 3.8	-1.9±10.4	-6.9, 3.1	-8.6±12	-14.4, -2.8	-6.2±1.8	-9.9, -2.4
PPGRF								
Knee flexion (-) / extension (+)	-50.2±7.8	-46.5, -54.0	-48.4±7	-45.0, -51.7	-41.1±12.3	-35.2, -47.0	-30.3±8.7	-26.2, -34.5
Peak Stance								
Knee flexion (-) / extension (+)	-53.7±7.1	-50.2, -57.1	-56.1±5.4	-53.5, -58.7	-58.6±9.3	-54.1, -63.1	-54.6±8.4	-50.6, -58.7
Knee abduction (-) / adduction (+)	-3.6±11	-8.9, 1.7	-4.8±8.6	-8.9, -.63	-14.4±8.2	-18.4, -10.5	-8.4±7.3	-4.8, -11.9
Hip flexion	50±9.8	45.3, 54.8	57.3±7.8	53.6, 61.1	58.2±12.4	52.3, 64.2	66.3±10.4	61.3, 71.3

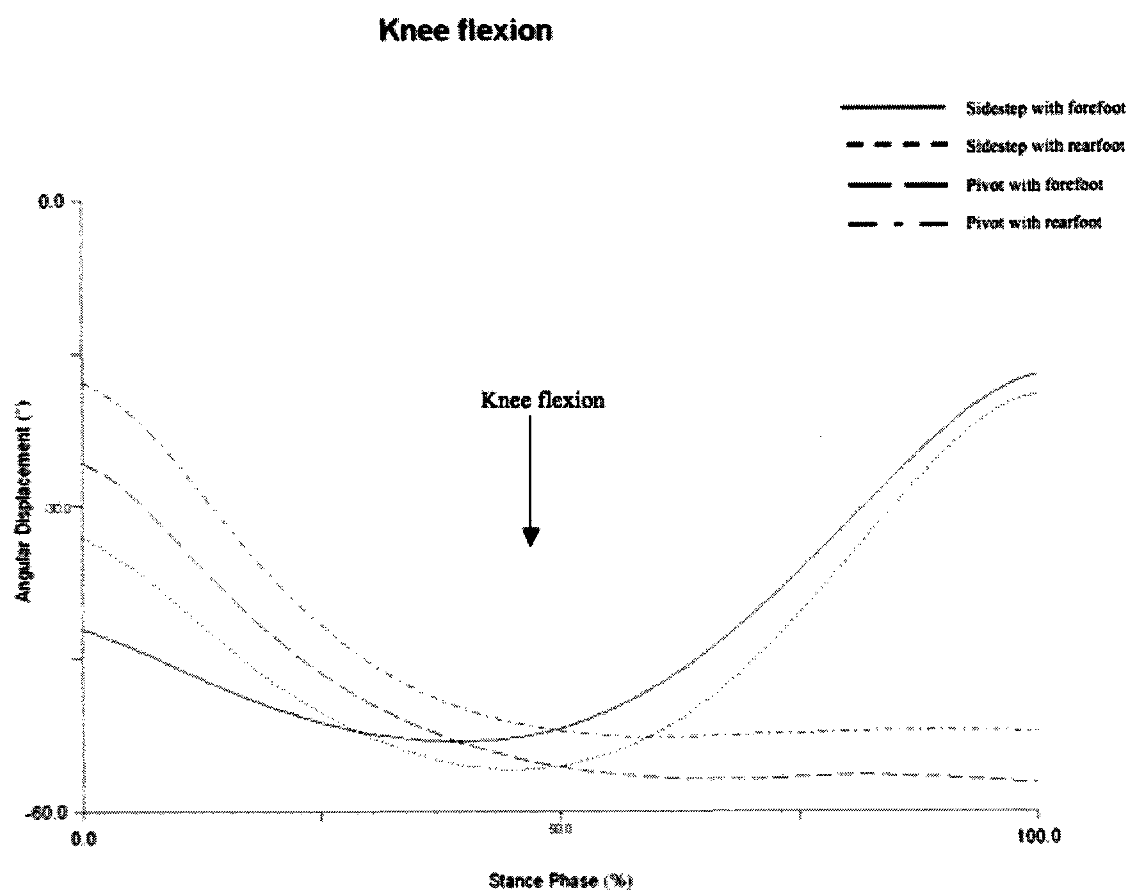
Appendix XIV - Table 5.2. Descriptive analysis (Means, SD, and 95% confidence intervals) of the kinetic variables at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PGRF), and peak stance during two athletic tasks using two landing-techniques

	Sidestep				**Pivot**			
	Forefoot		**Rearfoot**		**Forefoot**		**Rearfoot**	
	Mean ± SD	CI	Mean ± SD	CI	Mean ± SD	CI	Mean ± SD	CI
Initial contact								
Posterior Ground Reaction Force	.04±.02	.03, .05	-.01±.02	-.02, -.002	.02±.01	.01, .02	-.0005±.01	-.0076, .01
Knee flexion moment (-) / extension moment (+)	-.18±.11	-.23, -.12	-.16±.08	-.20, -.11	-.22±.08	-.27, -.18	-.14±.08	-.18, -.10
Knee abduction moment (-) / adduction moment (+)	.04±.06	.02, .07	.001±.03	-.01, .02	.10±.08	.06, .15	.011±.04	-.01, .03
Ankle plantarflexion moment (-) / Ddorsiflexion (+)	-.02±.02	-.01, -.03	.01±.01	-.001, .01	-.02±.01	-.03, -.02	-.02±.02	-.03, -.01
PVGRF								
Vertical ground reaction force	1.7±.4	1.5, 1.9	1.4±.4	1.3, 1.6	1.2±.7	.87, 1.57	1.4±.4	1.2, 1.6
PPGRF								
Posterior ground reaction force	.24±.12	.20, .31	.25±.11	.22, .31	.64±.42	.47, .87	.63±.23	.52, .76
Peak Stance								
Knee flexion moment (-) / extension moment (+)	1±.3	.87, 1.22	1.1±.3	.95, 1.2	.48±.2	.38, .58	.46±.2	.35, .56
Knee abduction moment (-) / adduction moment (+)	.34±.3	.20, .53	.41±.2	.30, .55	.61±.3	.46, .76	.62±.3	.46, .79

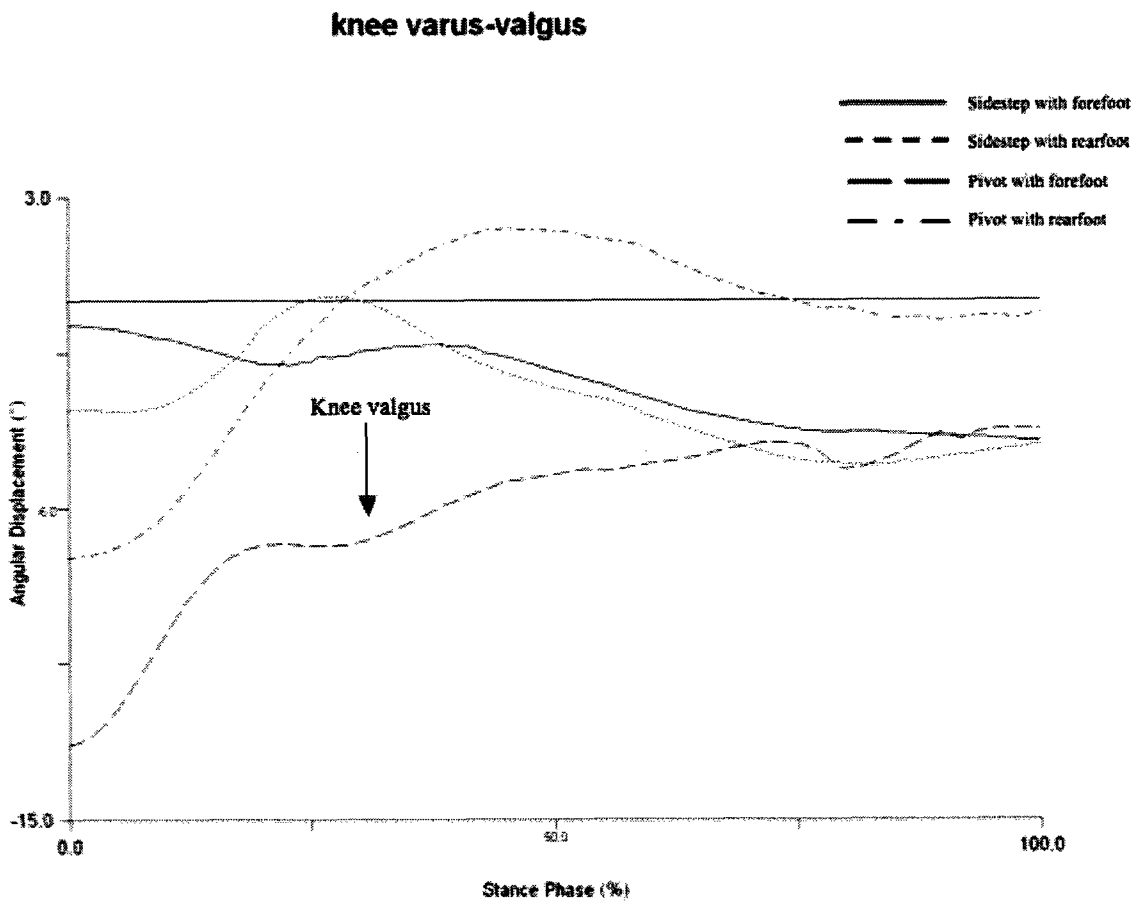
Appendix XV - Figure 5.1 Hip flexion angular displacement during two tasks performed with two landing techniques throughout the stance phase. Measured in degrees



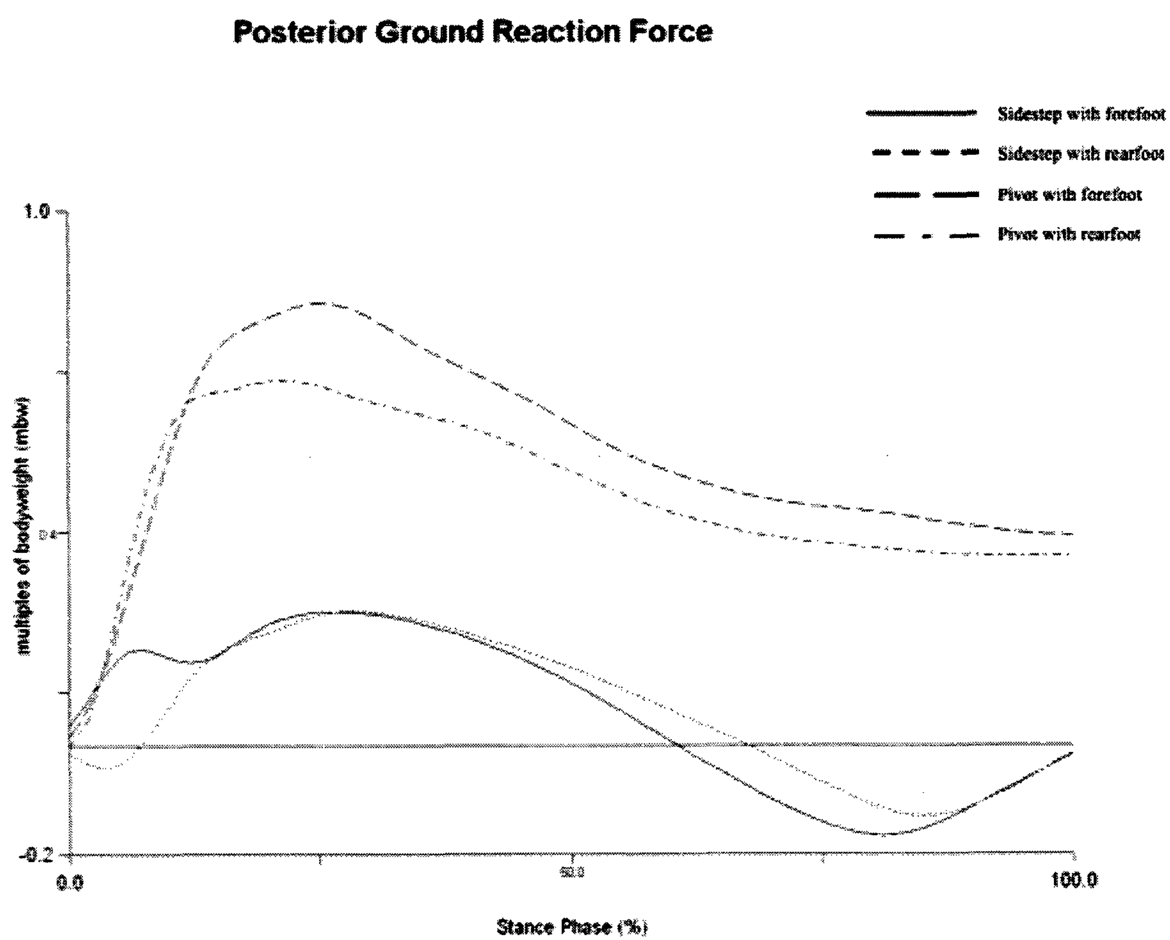
Appendix XVI - Figure 5.2 Knee flexion angular displacement during two tasks performed with two landing techniques throughout the stance phase. Measured in degrees



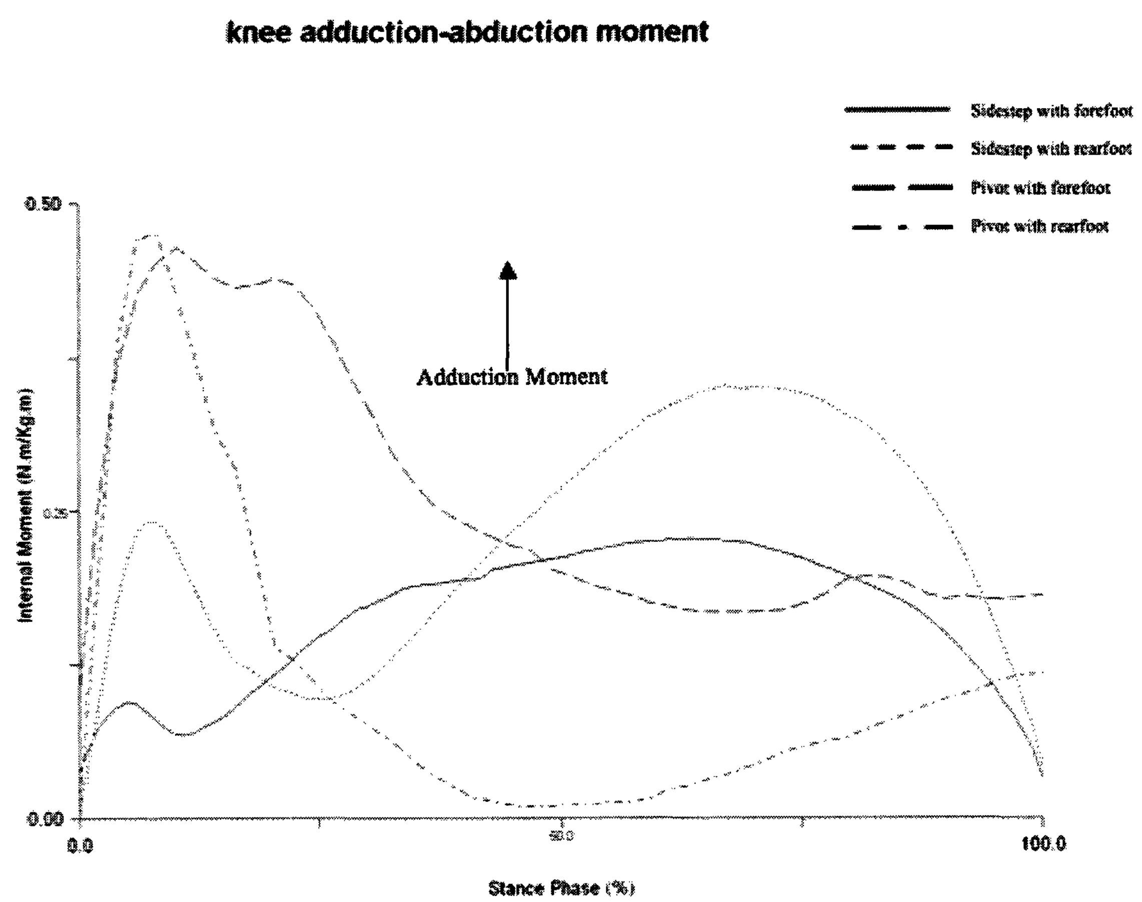
Appendix XVII - Figure 5.3 Knee valgus angular displacement during two tasks performed with two landing techniques during the stance phase. Measured in degrees



Appendix XVIII - Figure 5.4 Posterior Ground Reaction Force during two tasks performed with two landing techniques during the stance phase. Measured in multiples of bodyweight



Appendix XIX - Figure 5.5 Knee adduction-abduction moment during two tasks performed with two landing techniques during the stance phase. Measured in Nm/Kgm



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    - Delsys wireless and wired EMG system,
    - Noraxon wireless EMG,
    - VICON hardware/software,
    - Visual 3D software,
    - Intersense accelerometers, and
    - Coulborn tremor system.